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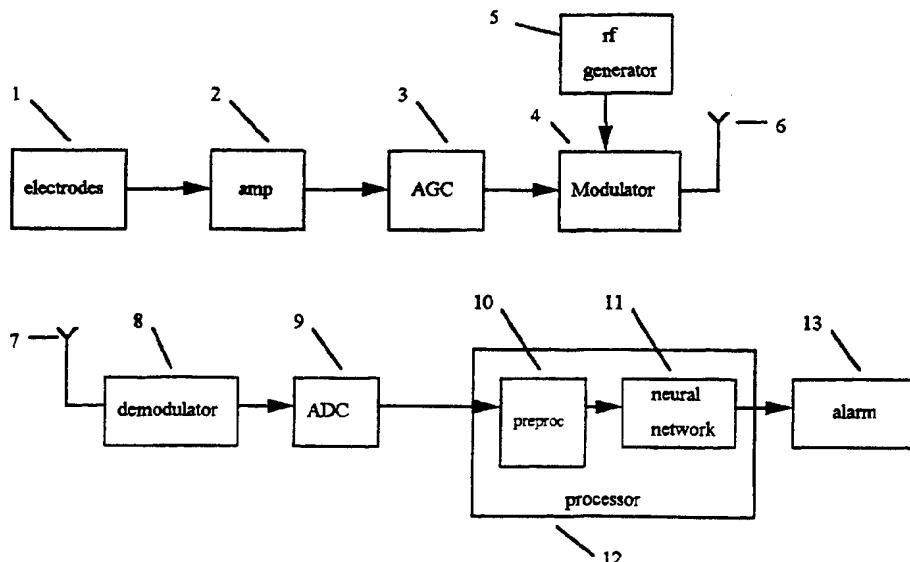
(57) Abstract

A heart monitoring apparatus and method is disclosed wherein an electrocardiograph signal is obtained from a patient and processed to enhance the salient features and to suppress noise. A plurality n of values representative of the features of the electrocardiograph signal are generated and used in a Kohonen neural network to generate an n dimensional vector. This vector is compared with a stored plurality m of n dimensional reference vectors defining an n dimensional Kohonen feature map to determine the proximity of the vector to the reference vectors. If it is determined by the Kohonen neural network that the vector is within or beyond a threshold range of the reference vectors a signal is output which can be used to initiate an event such as the generation of an alarm or the storage of ECG data.

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HEART MONITORING APPARATUS AND METHOD

The present invention generally relates to apparatus and a method for monitoring the operation of the heart of a patient. More specifically, the present invention relates to the analysis of electrocardiograph signals obtained from a patient using a neural network to monitor changes in the functioning or performance of the heart of a patient.

The electrical signals associated with muscular expansion and contraction of the heart and its chambers are frequently monitored to determine diseased conditions of the heart as well as deterioration and improvement of the condition of the heart, in particular following surgery or other treatment of the disease. The patients undergoing exercises or rehabilitation to improve the functioning of the heart.

Electrical signals of the heart are usually detected by means of conductive pads or contacts attached to the external chest wall and directly wired to a suitable machine which provides a graphical trace of the waveform, for a suitable display device on which the graphical wave form is represented for analysis by a suitably trained person. This however requires the patient to be permanently attached by wires to the monitoring apparatus and further requires the intervention of a suitably trained person.

2

Therefore an object of the present invention is to provide a heart monitoring apparatus and method which can monitor changes in heart condition automatically.

In accordance with one aspect the present invention provides heart monitoring apparatus comprising detection means for obtaining an electrocardiograph signal for a patient during a monitoring phase, preprocessing means for processing said electrocardiograph signal to enhance the salient features of the electrocardiograph signal and suppress the noise and to generate a plurality n of values representative of the features of said electrocardiograph signal, storage means for storing a plurality m of n dimensional reference vectors, Kohonen neural network means for receiving said plurality of values during the monitoring phase, for forming an end dimensional vector from said plurality of values, and for comparing said n dimensional vectors with said stored plurality m of n dimensional reference vectors defining an n dimension Kohonen feature map to determine the proximity of said n dimensional vector to said reference vectors, and output means for outputting a signal if said Kohonen neural network means determines that said n dimensional vector is within or beyond a threshold range of said reference vectors.

In accordance with a second aspect of the present invention there is provided a heart monitoring method comprising the steps of obtaining an electrocardiograph

3

signal from a patient during a monitoring phase, preprocessing the electrocardiograph signal to enhance the salient features of the electrocardiograph signal and suppress the noise, and to generate a plurality of n of

5   values representative of the features of the electrocardiograph signal, forming an n dimensional vector from the plurality of values, comparing the n dimensional vector with a stored plurality m of n dimensional reference vectors defining an n dimensional

10   Kohonen feature map to determine the proximity of the n dimension vector to said reference vectors, and outputting a signal if it is determined that the n dimensional vector is within or beyond a threshold range of said reference vectors.

15   In order for the Kohonen neural network means to generate the n dimensional Kohonen feature map, in accordance with one embodiment, during a learning phase, a plurality of reference values are input to the Kohonen neural network, which values are representative of

20   features of a reference electrocardiograph signal. The Kohonen neural network generates the n dimensional Kohonen feature map having the plurality m of reference vectors and these reference vectors are stored in the storage means.

25   The reference electrocardiograph signal can either be obtained from the patient during the learning phase to provide patient specific normal data, or the reference

4

electrocardiograph signal can be obtained from a large sample of the population to represent general normal heart operation. In either case the Kohonen neural network means operates during the monitoring phase to determine whether the n dimensional vector or the electrocardiograph signal lies outside a predetermined range of the reference vectors. If it is determined that the n dimensional vector lies outside the predetermined range an output signal is generated. This output signal can be used to trigger an event such as the generation of a warning, or the triggering of storage of the electrocardiograph signal either in an unprocessed condition or as the plurality of values. Also, the electrocardiograph signal could be stored in a compressed form other than as the plurality of values. Further, the output signal which can comprise an error vector can be used for further processing in order, for example to monitor the level of stress experienced by the heart.

In an alternative embodiment, the learning phase can be omitted and the plurality m of n dimensional reference vectors defining the n dimension Kohonen feature map can be input directly where these have been learnt previously on for instance another machine.

In another embodiment of the present invention which can be used for detecting specific heart conditions, at least one set of reference vectors can be built into the Kohonen feature map either by directly inputting the

5

vectors or by learning them from one or more sets of electrocardiograph signals representing known abnormalities. In this way, during the monitoring phase, when it is detecting by the Kohonen neural network means

5   that the n dimensional vector lies inside a predetermined range of the abnormal reference vectors, an output signal can be generated which can be used to trigger an alarm and/or trigger the storage of the electrocardiograph signal.

10   In one embodiment of the present invention the Kohonen neural network means is adapted to carry out the comparison and determination by determining the differences between the reference vectors and the plurality of values for each dimension, summing the

15   individual dimensional differences, and comparing the sum with a threshold difference value.

        In an alternative embodiment of the present invention the Kohonen neural network means is adapted to carry out the comparison and determination by mapping the

20   plurality of values onto the Kohonen feature map, calculating the differences therebetween by pythagoras law, and comparing the differences with at least one difference threshold.

        Therefore, in accordance with either of these

25   embodiments it is determined that the threshold has been exceeded, an output signal can be generated. This signal could simply be a trigger signal or can comprise the

6

error vector between the n dimensional vector and the nearest reference vector in the n dimensional Kohonen feature map. If the error vector is output, this can be used for further processing together with other heart
5   operation related signals in order to measure the level of stress experienced by the heart.

In one embodiment the further processing is carried out by a neural network such as the multi-layer perceptron. The other signals which can be input are for
10 example the heart rate, the heart rate variability, and other specific features related to the shape of the electrocardiograph signal.

In order to condition the electrocardiograph signals for input to the Kohonen neural network, they are pre-
15 processed in order to provide n values to form the n dimensional vector.

In one embodiment the n values are obtained by extracting n important features of the electrocardiograph signal e.g. by selecting peaks and other specific
20 features.

In an alternative embodiment of the present invention the electrocardiograph signal is transformed using a Fourier transform and/or a Wavelett transform.

Because an electrocardiograph trace can include
25 distinctive irregular heartbeats such as an ectopic beat, one embodiment of the present invention includes means to remove the distinctive irregular heartbeats from the

7

electrocardiograph signal which is processed by the Kohonen neural network to generate the output signal.

Since such irregular heartbeats can include distinctive frequency components e.g. a ventricular ectopic beat comprises a significant low frequency component, a high pass filter can be used to remove such beats and a low pass filter can be used to identify and indicate when such beats occur. In an alternative embodiment of the present invention the Kohonen neural network uses two different sets of reference vectors and operates in two steps to initially separate the distinctive irregular beats. A first set of n dimensional reference vectors are stored for identification of the distinctive irregular beats and a second set of n dimensional reference vectors are stored for monitoring regular heartbeats. During the monitoring phase the Kohonen neural network operates to initially compare the n dimensional vector with the stored first set of reference vectors to identify the distinctive irregular heartbeats, and to subsequently compare an n dimensional vector formed from regular heartbeats excluding the distinctive irregular heartbeats with the second set of n dimensional reference vectors. Thus in accordance with this method the Kohonen neural network can identify a distinctive irregular heartbeat by determining if the n dimensional vector falls within a threshold range of the first set of reference vectors.

8

The identified distinctive irregular heartbeats can then be ignored in the subsequent step carried out by the Kohonen network means and a count can be kept of their occurrence.

5   In addition to the removal of distinctive irregular heartbeats from the electrocardiograph signal, in order to increase the signal to noise ratio, the electrocardiograph signal or the plurality of values are preferably averaged over several regular heartbeats.

10   Also, to increase accuracy the average values are preferably normalised by reducing the values and dividing by the standard deviation.

In this way the electrocardiograph signals are preprocessed to remove distinctive irregular beats and

15   to significantly improve the signal to noise ratio thereby reducing the number of false identifications of novel electrocardiograph signals by the Kohonen neural network. The present invention can thus provide a heart monitoring apparatus which can monitor changes in heart

20   operation during the monitoring phase. The apparatus can be set to operate using either patient specific normal data or normal data representative of the population. Also, the apparatus can be sensitive to detect specific heart conditions which are represented by a specific and

25   distinctive electrocardiograph shape.

When the apparatus is used with patient's specific normal data, the normal range for heart operation can

9

simply be learnt during the learning phase such as whilst
a patient is still in hospital or still in intensive
care. Subsequently the functioning of the heart can be
compared with the normal patient specific range during
a monitoring phase e.g. following discharge from the
hospital or following a period of observation on a ward
away from intensive care. In this patient specific mode,
if the functioning of the heart is determined by the
Kohonen neural network as straying from within the normal
range learnt during the normal phase, then an event can
be triggered such as the raising of an alarm or the
storage of electrocardiograph data. The raising of the
alarm can allow for the intervention of a suitably
trained person. Whilst the alarm can indicate a
deterioration in the condition of the patient, it can
equally indicate an improvement in the condition of the
patient. Thus, in this patient specific mode, in either
case, if the normal range of operation of the patient's
heart has moved, the apparatus can be set into its
learning phase once more to relearn the patient's normal
heart operation range.

An important aspect in the present invention is that
in the present invention a change in functioning of the
heart is determined by considering a full analysis of the
electrocardiograph signal rather than by merely taking
into consideration the pulse rate. In the present
invention the Kohonen neural network is able to analysis

10

the shape of the electrocardiograph signal.

The advantage of having training data derived from the sample of the population is that some of the information contained within the electrocardiograph
5 signal is common across all patients. In addition, by having training data derived from the patient, the individual's unique features may also be classified. By obtaining both types of data the classification of electrocardiograph patterns is possible.

10 An embodiment of the present invention can comprise a self-contained battery operated portable unit incorporating a display and/or an audio output to indicate the presence of an alarm and/or the level of stress. Also, a connector for the transfer of classified
15 or stored data can be provided to allow the display or printing of data for a subsequent analysis by a suitably trained person. The present invention can also be embodied as a remote monitoring arrangement wherein the detection means and the output means are remote from one
20 another and communicate with one another by airborne communication e.g. radio frequency signals. This provides the advantage that the patient is not restricted by wires normally provided for an electrocardiograph apparatus and this arrangement is therefore ideally
25 suited for a wide range of applications.

The remote monitoring embodiment of the present invention may take many forms. The electrocardiograph

11

signals can be transmitted in either an analogue or digital form or indeed the module which is attached to the patient could carry out the processing and even some form of limited display and/or alarm. In one embodiment, only an output signal could be transmitted to a remote station.

Embodiments of the present invention will now be described with reference to the accompanying drawings, in which:-

Figure 1 is an example of an electrocardiograph trace;

Figure 2 is a schematic diagram of the heart monitoring apparatus in accordance with one embodiment of the present invention;

Figure 3 is a more detailed schematic diagram of the apparatus in accordance with one embodiment of the present invention;

Figure 4 is a more detailed schematic diagram of the apparatus in accordance with a second embodiment of the present invention;

Figure 5 is a more detailed schematic diagram of the apparatus in accordance with a third embodiment of the present invention;

Figure 6 is a more detailed schematic diagram of the apparatus in accordance with a fourth embodiment of the present invention;

Figure 7 is a more detailed schematic diagram of the

12

apparatus in accordance with a fifth embodiment of the present invention;

Figure 8 is a diagram illustrating the principles of wavelet transformation;

Figure 9A is a diagram of a two dimensional pattern space illustrating novelty detection;

Figure 9B illustrates alternative shapes from the regions around the reference vectors;

Figure 10 is a diagram of a two dimensional pattern space illustrating the principles of novelty detection; and

Figure 11 is a diagram of a two dimensional pattern space illustrating the principles of both novelty detection and the diagnosis of a specific heart condition;

Figure 12A is a schematic diagram of a transmission apparatus in accordance with a sixth embodiment of the present invention;

Figure 12B is a schematic diagram of the receiver apparatus in accordance with the sixth embodiment of the present invention;

Figures 13A and 13B are circuit diagrams of the transmitter apparatus illustrated in Figure 8A;

Figures 14A and 14B are circuit diagrams of the receiver apparatus for generating the serial analogue data in the receiver apparatus of Figure 8B;

Figure 15 is a diagram of the received

13

electrocardiograph signal;

Figure 16 is a diagram of the initial window of electrocardiograph data taken in the operation of the sixth embodiment of the present invention;

Figure 17 illustrates the method of identifying the R wave peak;

Figure 18 illustrates the method of centring a one second window around the R wave peak;

Figure 19 illustrates the method of determining the heart rate;

Figures 20a, 20b and 20c illustrate the identification of the twelve values from the electrocardiograph trace;

Figure 21 illustrates a display of the twelve values obtained;

Figures 22a and 22b are flow diagrams illustrating the operation of the sixth embodiment;

Figure 23 is a flow diagram illustrating the feature extraction step; and

Figure 24 is a schematic diagram of a seventh embodiment of the present invention.

Referring now to the drawings, Figure 1 illustrates a typical electrocardiograph trace wherein various features P, Q, R, S and T can be seen. The shape and size of each of the features is an important indication of the condition and operation of the heart. The shape of the electrocardiograph trace for any two patients is

14

never exactly the same and therefore detecting differences and specific heart conditions for patients requires significant skill by a medical practitioner. The present invention is directed towards obtaining

5   information on the shape of the electrocardiograph trace to enable an automatic detection of a change.

Figure 2 illustrates a highly schematic remote heart monitoring apparatus in accordance with one embodiment of the present invention. A patient 100 is fitted with

10 a portable housing 200 which can obtain an electrocardiograph signal from the patient 100 and transmit this to the remote base station 600. The electrocardiograph signal can be obtained from the patient using conventional electrodes such as an

15 arrangement of three electrodes in line with existing medical practices. Two of the electrodes can detect a differential signal whilst the third can provide a common or reference signal. The differentially detecting signal can then be transmitted to the remote base station 600

20 for analysis. As will be seen hereinafter in accordance with another embodiment of the present invention, the analysis of the electrocardiograph signal can also take place within the portable housing 200 carried by the patient.

25 Conveniently, the portable housing 200 can be carried by the patient 100 by placing it within a pocket in a vest or similar garment so that the electrodes are

15

held against the patient's skin at the correct position.
Alternatively, the electrodes are provided separate from
the housing and connected thereto by wires.   At the
remote   base   station   600   the   transmitted
5   electrocardiograph signal is received by a receiver 300
and passed to a neural computer 400 to processing. The
neural   computer   400   analyses   the   electrocardiograph
signal and passes the signal to a display and audible
alarm device 500 if a novel electrocardiograph signal is
10   detected. A neural computer is also capable of storing
the   electrocardiograph   signal   for   the   conventional
analysis by a skilled medical practitioner.

Figure 3 illustrates in more detail one embodiment
of the present invention wherein the electrocardiograph
15   signal is transmitted as a radio frequency signal. The
electrocardiograph signal detected by the electrodes 1
is   passed   to   an   instrument   amplifier   2   having
characteristics of very high input impedance, very low
common mode signal rejection and forward gain of 1000
20   over a bandwidth of 0.01Hz to 500Hz. The output of the
amplifier 2 is then fed to a very slow response automatic
gain control circuit 3 to ensure that the peak output
signal level is kept reasonably constant at the maximum
permitted   modulation   level   without   introducing
25   distortion. The signal is then modulated in a modulated
form with an RF signal generated by the RF generator 5.
AM or FM modulation may be used as appropriate. The

16

signal is then passed to an antenna 6 for transmission. The radio transmitter is a low power short range circuit designed for operation within a building or hospital ward, and with a frequency stability such that multiple

5   channels may operate within the frequency band.   At the base station, a radio receiver comprising an antenna 7 and a demodulator 8 receive and demodulate the cardiac signal, one per channel, each sensitive enough to detect the low power transmitted signal and selective enough to

10   discriminate   between   adjacent   signals   without interference. The cardiac signal is then converted to a digital signal by means of a linear analogue to digital converter 9 at a sampling rate of 500Hz, and resolving its sample of the analogue signal into a binary number

15   between and twelve and sixteen bits in length. The binary code signal is then passed to the processor 12 for pre-processing within the pre-processor unit 10. The pre-processor signals are then passed to the neural network   11   for   analysis   and   if   a   normal

20   electrocardiograph trace is determined an output is triggered which in the current example causes an alarm unit 13 to raise an audible and/or visual alarm. In addition or instead, the output can trigger storage and possibly the display of the electrocardiograph trace.

25   Figure 4 illustrates a second embodiment of the present invention wherein an electrocardiograph signal detected   by   the   electrodes   1   is   passed   to   an

17

instrumentation amplifier 2 and automatic gain control circuit 3 as for the previous embodiment. However, before onward transmission to the radio transmitter, the signal is digitised by an analogue to digital converter
5   14 to form a binary bit stream which is then used to modulate the RF signal from the RF generator 5 in the modulator 4.

At the base station, the receiver comprises the antenna 7 and the demodulator 8 demodulates the received
10 signal to provide a digital signal which is a digitised sample of the original analogue signal. This signal is then passed into the processor 12 and is analyzed in the same manner as for the previous embodiment.

Referring now to Figure 5 there is illustrated a
15 third embodiment of the present invention wherein the electrocardiograph signal from the electrodes is directly processed by a high resolution analogue digital converter 15 capable of resolving to better than one micro-volt. The digitised signal is then processed using a digital
20 signal processor 16 to normalise the amplitude, improve the signal to noise level and bandwidth limit the signal. The digital data from the digital signal processor 16 is then used to modulate an RF signal from the RF generator 5 using the modulator 4.

25 At the base station the RF signal is demodulated and the serial digital signal is then converted to the parallel in unit 17. The parallel signal comprises

18

twelve to sixteen bit binary coded words which are
presented to the processor 12 at a rate of 500Hz. The
processor 12 then operates on the received signal in the
same manner as for the previous embodiments.

5   Figure 6 illustrates a fourth embodiment of the
present invention which is similar to the previous
embodiment of the present invention except that the
analysis of the electrocardiograph signal takes place
within the portable housing carried by the patient. The
10   electrocardiograph signal from the electrodes 1 is passed
to a high resolution analogue to digital converter 15.
The output of the high resolution analogue to digital
converter 15 is then passed to a digital signal processor
16 to normalise the amplitude, improve the signal to
15   noise level and bandwidth limit the signal. The output
of the digital signal processor 16 is then passed to the
processor 12 wherein the electrocardiograph signal is
first pre-processed in the pre-processing unit 10. The
pre-processed data is then passed to the neural network
20   11 for analysis. If a novel electrocardiograph trace is
determined, an output signal is generated which is in the
present embodiment an alarm signal. This signal is then
passed to the modulator 4 for modulating the RF signal
from the RF generator 5. The modulated RF signal is then
25   transmitted by the antenna 6. At the base station, the
antenna 7 receives the modulated RF signal which is then
demodulated in the demodulator 8. The alarm signal is

19

then passed to an alarm unit 21 to generate an audible
or visual alarm.

In addition to the alarm unit, the neural network
11 can output summary data of the cardiac signal such as
5   heart rate.   The alarm unit 21 can also receive such
summary data for display.

Figure 7 illustrates a fifth embodiment of the
present invention which is similar to the embodiment of
Figure 6 except that the signals are transmitted in
10   analogue form.   The output of the neural network 11 is
passed to a digital to analogue converter 22.   The
analogue output of the digital to analogue converter 22
is then used for modulation of the RF signal from the RF
generator 5 in the modulator 4.

15   At the base station the analogue output of the
demodulator 8 is digitised using an analogue to digital
converter 22 and the digital signal is then passed to the
alarm unit 21.   In embodiments 1 to 5 described
hereinabove the raw electrocardiograph signal itself is
20   not suitable for immediate processing by the neural
computer. Small variations in the signal are significant
and need to be detected, but when embedded in the whole
signal they become numerically insignificant compared
with the noise level, thus making it virtually impossible
25   to carry out novelty detection on the raw signal.   To
overcome this, digital signal processing techniques are
used to extract information from the received signal.

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Extracted information comprises significant features in the electrocardiograph signal which are measured and used by the neural computer to detect novelty.

The features that are important in an electrocardiograph signal are peak heights and the separation between peaks. One method of pre-processing the electrocardiograph signal is to utilize the fast Fourier transform (FFT). The fast Fourier transform obtains frequency information. Another or complementary technique is to use the Wavelett transform. Alternatively, as will be described hereinafter with reference to the sixth embodiment, features can be extracted from the electrocardiograph signal.

In one embodiment the electrocardiograph signal is sampled at the frequency of 50Hz typically a sample of 4 seconds is used which will typically provide at least five complete heartbeat signals within the window. Preparing this data for the fast Fourier transform (FFT) can take a number of forms.

1.   First 128 samples can be presented to a standard FFT algorithm to obtain 64 real and 64 imaginary components.

2.   A peak search can be carried out in the first quarter of the data to find the highest amplitude signal. This can be assumed to be a capital R point on the electrocardiograph. The following 128 samples can be presented to a standard FFT algorithm to obtain 64 and

21

64 imaginary components. This has the advantage of synchronising the data and resulting in a less noisy FFT.

3. The peak search can be carried out in the second quarter of the data to find the highest amplitude

5   sample. This can be assumed to be an R point on the electrocardiograph. 128 samples can be selected from a fixed number of samples preceding the peak. These samples can then be presented to the standard FFT algorithm to obtain 64 real and 64 imaginary components.

10 This has the advantage over the second method given above of reducing the edge effect caused by starting the sample with a peak.

Any of the above methods can be enhanced with the use of apodisation, i.e. an edge effect reducing

15 algorithm such as a hanning window.

More than one pulse is included in the sample presented to the FFT to allow pulse rate information to be taken into account. An alternative embodiment can use a single electrocardiograph pulse for pre-processing and

20 a separate pulse rate algorithm to calculate the pulse rate. This may make the FFT more sensitive to subtle changes in the electrocardiograph waveform.

Preferably only the real components of the fast Fourier transform are used in the Kohonen feature map.

25 However, the imaginary components or a power spectrum can also be used instead of or in addition to the real components.

22

In addition to, or instead of, the fast Fourier transform, Wavelett transforms can be used to pre-process the raw electrocardiograph signal. This transform operates by measuring the similarity of the signal to a

5   set of defined wave functions. Such a set of wave functions are illustrated in Figure 8. The wave functions are arranged in a hierarchy so that output is scaler in nature. The transform can provide typically 64 values.

10   As with the fast Fourier transform, multiple pulses could be presented or a single pulse with additional pulse rate information given separately. The Wavelett transform method also requires that the signal is synchronised.

15   In one embodiment all of the values produced by the fast Fourier transform and the Wavelett transform are used as inputs to the Kohonen feature map. A system using the fast Fourier transform with a multiple pulse window provides 64 values which are the real components

20   of the fast Fourier transform and which forms a vector having 64 values for the Kohonen feature map. Where both the fast Fourier transform and the Wavelett transform are used for a single pulse window, 128 values, comprising 64 fast Fourier transform values and 64 Wavelett

25   transform values are provided.

The neural computer operates as a Kohonen feature map. Such a neural network is well known and reference

23

to this can be found in 'Neural Network Architectures.
An introduction', J Dayhoff, Van Nostrand, Reinhold 1990,
pages 163 to 191.

Feature mapping in the Kohonen feature map is a
process by which example training vectors are clustered
in featured space. The pre-processed values from an
electrocardiograph signal define a vector position in an
Euclidean multi-dimensional space. The dimensionality
of the space is determined by the number of features
measured i.e. the number of values. For example if two
features are measured with values of 5 and 3 the vector
position for that electrocardiograph is (5,3) in a two
dimensional space. In the embodiments of the present
invention described hereinabove, where 64 FFT values are
provided, the feature space is a 64 dimensional space.
For a more complex embodiment using the fast Fourier
transform output and the Wavelett transform output the
Kohonen feature map has 128 dimensions.

In the Kohonen feature map a plurality of reference
vectors are used which are initially randomly positioned
in the multi-dimensional space. During a learning phase
when values are input to the Kohonen feature map, the
position of the reference vectors are moved to represent
the distribution of the values in the feature space.
Each of the reference vectors has a point in the multi-
dimensional space which is its centre and can have a
predefined area of influence therearound. This is

24

illustrated in Figure 9A. The area surrounding the
reference vector defines a predetermined distance in
pattern space from the reference vector. As can be seen
in Figure 9B, the shape of the area of influence can be
5   varied between radial, square and diamond, respectively.
The radial area of influence is the most accurate but is
the most complex to compute.

Figure 9A illustrates a single reference vector is
a two dimensional space. Surrounding the reference
10   vector is the predetermined region which is considered
to be within the range of the reference vector. If the
reference vector were to represent a learnt reference
vector or an electrocardiograph signal, when an
electrocardiograph value is input in the monitoring
15   phase, if it falls within the area of influence it is
recognised whereas if it falls outside the area of
influence it is not recognised and there is therefore
considered to be novelty: resulting in the generation of
an output signal which can for instance trigger an alarm
20   or trigger the storage of the electrocardiograph signal.
Also, the error vector which is the vector between the
nearest reference vector and the vector of the input data
can be output.

Considering for simplicity the two dimensional
25   pattern of space, if a reference vector has a position
(5, 3) and the electrocardiograph signal obtained during
a measuring phase has a vector position of (6, 2.5), the

25

difference between the two can be calculated either by pythagoras' law or simply by summing individual dimensional differences: which in this case gives the difference of 1.5.

Feature mapping defines reference positions in the feature space which describe "normal" positions for an electrocardiograph signal. A threshold can be set around each reference vector which describes the limit of the vectors represented in space. Thus the difference of, for example, 2.2 can be defined as no longer represented by the "normal" vector and there is thus novelty. This is illustrated in Figure 9A.

The position of the reference vectors is determined during a training process which is described in more detail in the book by J Dayhoff. The algorithm takes the form of a neural network which each neuron network defines a reference vector position. Given the sample data, an iterative learning process is used to find the optimum position for all of the reference vectors based on the position and density of the example data. This is the networks training process.

Figure 10 illustrates a simple case of a two dimensional pattern space having eight reference vectors, each with their own area of influence. The reference values represent optimised mean values derived from the electrocardiograph values during the learning phase and are placed according to the distribution and density

26

within the pattern space. The periphery of the area of influence for each of the reference vectors therefore defines a threshold of normality for the electrocardiograph signal based on the electrocardiograph signals during the learning phase.

Figure 10 illustrates two electrocardiograph values for this two dimensional hypothetical example: one representing a normal electrocardiograph signal which falls within the threshold of normality and one representing a novel electrocardiograph signal which falls outside the threshold of normality.

In one embodiment of the present invention where for example a patient is being monitored in a hospital environment, example data is collected from the patient used to train the network. In an alternative embodiment training data can be obtained from electrocardiograph signals from a large sample of the population. A set of reference vectors will be defined and the reference vectors collected from the patient will be unique to that patient. During the monitoring phase the pre-processed electrocardiograph values are input to the neural computer to determine the difference between the reference vectors either by using Pythagoras' law or simply by summing the individual dimension differences. Alternatively, a radial basis function algorithm can be used. In this case the centres and widths of the functions are defined by a similar training process as

27

above.   Each centre has a region of influence which may
be radial, square or binary shaped, as shown in Figures
9A and 9B.   When an input is presented to the patent
space if it falls inside at least one centre region it
5   is recognised, otherwise it is treated as a novelty.

Figure 11 illustrates a further embodiment of the
present invention which provides for some diagnostic
capability.   In this embodiment specific reference
vectors are defined representing specific conditions such
10   as myocardial infarction.   This region in pattern space
defines a specific heart condition and can be obtained
by inputting electrocardiograph signals which are known
to represent the operation of a heart during the specific
condition.   In this embodiment during the monitoring
15   phase, if the electrocardiograph signal values fall
outside the threshold of normality and inside the region
of abnormality, the apparatus can give an indication of
the specific heart condition which has arisen.

Whilst Figure 11 illustrates only a single region
20   in pattern space representing a single specific heart
condition, clearly this approach can be used to define
many different regions in pattern space representing many
different specific heart conditions.

The neural computer is thus capable of generating
25   a signal which can cause an alarm to be raised either
audibly or visually.   One embodiment of the present
invention also provides the capability to equal the

28

audible or visual warning device to give an indication
of the specific heart condition which has arisen if this
has been recognised by the neural computer.

Referring now to Figures 12A and 12B, a sixth
5 embodiment of the present invention will now be
described. Figure 12A illustrates a transmitter module
which can be carried by a patient. Conventional
electrocardiograph signals LA, LG, RA are received by the
instrumentation amplifier 25 which amplifies the
10 differential signal and outputs it to an anti-alias
filter 26 to move high frequencies which can cause
sampling errors. The filtered signal is then passed on
to an analogue to digital converter 27 and the digitised
signal is then input to a serial shift register 28. The
15 analogue to digital converter 27 and the serial shift
register 28 are under the control of the frame timing and
control unit 30. The output of the serial shift register
28 is then input to the RF modulator and transmitter 29
for transmission of the electrocardiograph signal on the
20 antenna 31 to the remote space station.

Figure 12B schematically illustrates the remote base
station. The transmitted electrocardiograph signal is
picked up by the antenna 32 and passed to the RF receiver
and demodulator 33. The demodulated electrocardiograph
25 signal is converted to a serial analogue signal by the
serial digital to analogue converter 34 under the control
of the frame timing and control unit 35. The serial

29

analogue signal is then input to a input/output card 37
within the computer 36. In this embodiment the computer
is an IBM (trademark) compatible computer and the
input/output card is the Lab-PC+ multiple channel
5   analogue and digital I/OCARD available from National
Instruments. The data acquisitional software used is the
NI/DAQ software available from National Instruments.

A description will now be made of the circuits
forming the transmitter and receiver units illustrated
10   in Figures 12A and 12B.

Figures 13A and 13B illustrate the circuit diagram
of the instrumentation amplifier 25, the anti-aliasing
filter 26, the analogue to digital converter 27, the
serial shift register 28, and the frame timing and
15   control unit 30 of Figure 8.

The electrodes placed on the patient detect a
differential signal of amplitude approximately one
millivolt. This signal is then passed via capacitors C5
and C6 and remove any DC component and then through a low
20   pass filter comprising resistors R7, R8 and capacitor C7.
DC restoration of the incoming signal is assured by
providing 1Mō leakage resistance R3, R4, R5 and R6 to the
earth point. Very high input impedance instrumentation
amplifier IC10 having again set to 2000 is used to
25   amplifier the electrocardiac signal. The output of IC10
is passed through a unity gain low pass filter IC12 which
has a roll-off frequency less than 100Hz. The output of

30

the low pass filter IC12 is input to a 12 bit analogue
to digital converter IC9. The analogue to digital
converter is configured to give a serial digital output
of 12 bit accuracy but only the ten most significant bits

5   of data are used.

The analogue to digital conversion cycle commences
when the chip select (CS) signal of pin 20 is taken low.
Immediately following this the analogue to digital
converter takes pin 9 low indicating it is busy. When

10   the conversion is complete the busy signal will be taken
high, and via a J-K flip-flop IC6. The (CS) signal will
also be taken high.

As each stage of the successive approximation
analogue to digital conversion occurs, a data bit and a

15   clock edge is generated by the analogue to digital
converter with the most significant bit first. These are
output on pins 18 and 17 respectively. The serial data
(SerDa) and clock (shIn) are routed to a shift register
IC5 and the data is clocked into the shift register at

20   the HiCk rate of 1.5 microseconds per bit.

The frame timing and logic control comprises four
main blocks: the oscillator and divider, the counter,
the gating logic, and the shift register. The oscillator
and divider comprises three inverters IC2A, IC2C and IC2D

25   connected in series together with a resistor R1 and a
capacitor C1 is connected ground between inverters IC2A
and IC2D. This forms an oscillator running at 667KHz

31

which is the HiCk frequency. The output of inverter IC2A
is fed to the asynchronous binary divider IC1 which
divides HiCk by 256 to produce a low frequency clock LoCk
running at 2.6kHz. A D-type flip-flop IC7A is used to
5   synchronise the output of the divider to HiCk.

The counter comprises a synchronous programmable
counter IC3 set to divide by thirteen. It does this by
being preset to three and then counting to fifteen
whereupon the carry out once began presets the counter
10   to three. This creates the data frame rate of 200 frames
per second from the 2600Hz clock.

The start of each frame is indicated by a Sync pulse
LoCk beats wide. This is generated by a J-K flip-flop
IC6A. When the output of Q3 of the counter IC3 goes
15   high, on the count transition from three to four, the J
input of the J-K flip-flop IC6A is taken high and on the
next LoCk rising edge, the Q output Sync of the J-K flip-
flop IC6A goes high. Two counts later the Q2 output of
the counter IC3 goes high as the counter goes from five
20   to six and the K input of the J-K flip-flop IC6A goes
high. On the next rising edge of LoCk the Q output of
the J-K flip-flop clocks back low. To prevent the J-K
flip-flop IC6A from toggling on the next LoCk edge, the
J-K flip-flop IC6A is held reset when the output Q4 of
25   the counter IC3 goes high. Sync is used to start an
analogue to digital conversion by having a leading edge
detector, comprising an inverter IC4C and a J-K flip-flop

32

IC7B generate a one beat pulse which puts the J-K flip-flop IC6B high and CS low to start the analogue to digital conversion. A set-reset flip-flop is created from the inverters IC4A and IC4B to provide gating for ten beats of LoCk by setting the flip-flop when the Q4 output counter IC3 goes high, count eight, and resetting the flip-flop low when the Sync goes high, count three. The output, clock enable (CkEn) is used to gate the clock at the shift register IC5.

The shift register IC5 is configured as a twelve bit shift register. The analogue to digital converter IC9 generates the twelve clock edges at 1.6 microsecond intervals as it completes a conversion and each data bit is clocked into the shift register IC5 using this. The data is clocked out at the much slower rate of 2600Hz. LoCk is gated with CkEn by the inverter IC8B to shift out the data in the shift register IC5. The serial data stream is mixed with the Sync pulse on the inverter IC8D and is gated with inverted LoCk from the inverter IC4D so that half width data pulses are generated in the serial output stream. Only ten LoCk edges are gated into the shift register IC5 so that the least significant two bits of data are ignored.

In this embodiment a proprietary low powered radio module is used for data transmission. This operates at 433MHz or 458MHz and operates FM. Modulation band width is in the region of 4000 baud and so is capable of

33

handling 2600 baud data rate in this particular configuration.

The circuit has been implemented as a low power design so that it can operate from low capacity batteries.

Referring now to Figures 14A and 14B, the receiver and decoder circuit comprises a proprietary FM receiver with two outputs, a signal out and a received signal strength indicator (RSSI). If RSSI is not present the decoder circuit is held reset. The RSSI signal is input to a diode D1 to provide a threshold level for the cut-off as the strength falls but before data corruption can occur. The transistor Q1 is used to switch the signal into the submit trigger IC2A to produce a valid data signal ValiDa. A green LED D3 is used to indicate the presence of this signal. The valid data signal is also used to gate the serial data so that only valid data is accepted. The transistor Q2 is a Sync detector. As long as data pulses are present capacitor C1 will be held discharged. Whenever a Sync pulse occurs which is two beats wide and therefore longer than any other pulse, the voltage across the capacitor C1 rises to the threshold of the schmitt trigger IC2C and (SyncDet) goes low. The signal goes high on the trailing edge of Sync.

The (SyncDet) signal is input to monostables IC7A and IC7B. The monostable IC7A is triggered on the leading edge of (SyncDet) and monostable IC7B is

34

triggered on the training edge of (SyncDet). The
monostables IC4A and IC4B are cross-coupled to form an
oscillator. This technique is used in order to overcome
any problems associated with oscillators where the first
5   few cycles after being enabled will have a longer
interval. With monostables the period of each cycle is
accurately determined. The oscillator comprising the
cross-coupled monostable IC4A and IC4B is triggered from
the output of the monostable IC7B. The preset counter
10   IC1A counts 12 cycles from the oscillator and then stops.
It turns off the clock by gating off monostable IC4B.
It is reset by IC7A and IC7B triggers monostable IC4A to
start the clock.

The valid serial data is gated by schmitt trigger
15   IC2B and IC2D and is fed to three and four bit shift
registers IC5A, IC5B and IC8A which are connected to give
twelve bits of serial register. Twelve clock cycles
occur shifting in ten bits of data and two bits of Sync.
The Sync bits are ignored and the bottom two least
20   significant bits of the shift register are not used. The
data is serially shifted in on one frame and on the
(SyncDet) of the next frame the output buffers IC3, IC6
and IC9 are updated with data. Ten bits of data are
available for input to the computer which can then use
25   this data to perform data extraction and novelty
detection. In the present embodiment this parallel data
is not used.

35

Instead of using the parallel data a serial analogue output is used from the serial to analogue converter 10 and low pass filter connected amplifier IC11A. The serial digital to analogue converter provides a 0 to 5 volt output for serially clocked-in binary data wherein the data is clocked by T1. (SyncDet) is used as the load or update signal. The amplifier IC11A is used as a low pass filter to smooth the analogue signal.

Thus, in this embodiment electrocardiac signals are transmitted from the portable patient module to the remote base station for analysis. Reference will now be made to the method of analysis carried out by the sixth embodiment of the present invention with reference to Figures 15 to 23. This is carried out by a program operating in the PC.

Figure 22A and 22B illustrate the method of analysis of this embodiment of the present invention. In step S1 electrocardiograph data is input in the form illustrated in Figure 15. The data is captured at a sample rate of 1KHz and stored as a binary *.WAV file in step S2. The binary *.WAV file is converted to an Excel (trademark) file in step S3 and in step S4 the amplitude information is extracted. In step S5 the data is sub-sampled at frequency of 100Hz. In step S6 a three second window ie. 300 samples is taken as shown in Figure 16. In step S7 the middle two seconds are scanned for an R wave peak. The peak is identified as shown in Figure 17 and in step

36

S8 a one second window is created with the R wave centralised as shown in Figure 18. In step S9 the entire three second window is then normalised such that the one second window has amplitudes ranging from 0 to 1. The next step as illustrated in Figure 22B is step S10 which is feature extraction. Figure 23 illustrates the steps in the feature extraction process. In step S100 the T wave is identified by finding the highest point within a window of preset distance from the R wave. This distance is defined by a priori information about electrocardiographs. In step S110 the amplitudes of both the S and R waves are then measure and in step S120 the mean value of the two is taken and used in step S130 to locate the next or previous R wave in the three second window by looking for a peak above the calculated average value. The larger section of the three second window to either side of the initially located R wave is scanned from the R wave to locate the next point at which the amplitude is higher than the mean value. A window is then placed around this location to locate the next R wave. Figure 19 illustrates this process.

From the measured interval between R waves, in step S140 the heart rate in beats per minute is calculated. The value calculated is a discreet value which is not smoothed by averaging over time. In step S150 the R to R variability i.e. the variation in heart rate is calculated for the most recent 200 heartbeats and this

37

is output together with the heart rate measurement.

The method then enters a series of steps designed to identify that a heartbeat shaped pattern is present and to locate irregular beats such as a ventricular ectopic beat or an artifact. A ventricular ectopic beat occurs when the conduction path through the heart is initiated from a rogue node in the ventricles. It is important to locate such beats for heart stress diagnosis, but must be considered as spurious with regards to monitoring heart conditions.

The main concept behind this part of the method is to transfer the signal from the time dimension to a dimension offering greater separation of signal types. This is achieved in this particular embodiment by segmentation of the signal. However, transforms such as the Fourier transform or more particularly the Wavelett transform may be employed in order to generate the requirement values.

In step S160 the positions of the PQ and S waves are calculated using a priori knowledge of beat shape. The midpoint of the P and Q waves occur during what is termed the P - R interval. This may be considered as a base line from which to measure amplitudes. The midpoint of the S and T waves has diagnostic important with regard to the recovery period of the ventricles. Thus in step S170 the midpoints PQ and ST are determined as shown in Figure 20A. Figure 20B shows the calculation of the

38

amplitude of these points and in Figure 20C the calculation of twelve values carried out in step S180 is illustrated.   Figure 21 illustrates the twelve values which can be obtained from an ECG trace.

5   The twelve points which are taken to analyze the wave form are

$$| P_{amplitude} - PQ_{amplitude} |$$

$$| PQ_{amplitude} - Q_{amplitude} |$$

$$| PQ_{amplitude} - ST_{amplitude} |$$

$$| PQ_{amplitude} - R_{amplitude} |$$

$$| PQ_{amplitude} - S_{amplitude} |$$

$$| PQ_{amplitude} - T_{amplitude} |$$

$$| P_{time} - Q_{time} |$$

$$| P_{time} - T_{time} |$$

$$| Q_{time} - R_{time} |$$

$$| Q_{time} - S_{time} |$$

$$| Q_{time} - T_{time} |$$

$$| S_{time} - T_{time} |$$

Having identified the twelve values representing the
20 electrocardiograph signal, at step S11 it is determined whether the learning or monitoring mode is required. If the learning mode is required to determine the reference vectors or positions of the neurons in the Kohonen neural network, the learning phase is initiated in step S12 by
25 the defining of the number and initial positions of the neurones. In step S13 the twelve values are then input

39

into the Kohonen neural network which in step S14 adaptively learns the positions of the neurones. This will continue for each of the three second windows of the input ECG data identified in step S6.

If in step S11 it is determined that monitoring is required, the twelve values are input into the Kohonen neural network to identify ectopic beats in step S15. The principle of operation of the Kohonen neural network in both the learning and monitoring modes is described in the book by J Dayhoff. In step S16 the nearest neuron (reference vector) is determined and in step S17 the error vector between the input vector and the nearest reference vector is determined. In step S18 it is determined whether the error is above the predetermined threshold. If the threshold is exceeded in step S19 the ectopic rate is calculated and output and in step S20 the beat is marked as an ectopic beat so that it is ignored in the further processing of the electrocardiograph signal. The processing can then continue as if the threshold had not been exceeded in step S18 and in step S21 the eight most recent one second sub windows which do not include an ectopic beat are averaged by median averaging. In step S22 the feature extraction is once again carried out as shown in Figure 23. In step S23 the twelve values are normalised using a pre-calculated mean and standard deviation for each value. The mean and standard deviation for each dimension is calculated for

40

an arbitrary 100 data sets and each value is reduced by the mean value and divided by the standard deviation in order to normalise the values.

In step S24 the ST segment depression has calculated and output. The ST segment depression holds diagnostic information with regards to the recovery ability of the ventricles. Thus, ST segment depression can be used as an indication of overall heart stress level.

The twelve normalised values are then input into the Kohonen neural network in step S25. The Kohonen neural network has neurons which were learnt during the learning phase (steps S12, S13 and S14). In step S26 the nearest neuron to the input vector is determined and in step S27 the error vectors are determined. In step S28 it is determined whether the threshold is exceeded and if so an output signal is generated in step S29 indicating novelty. In step S29 also the error vector determining step S27 can be output.

If the threshold is not exceeded once the signal has been output in step S29 it has been determined that the threshold has been exceeded, the next three second window can be analyzed i.e. the process can return to step S6. In the method of operation of the sixth embodiment five outputs are generated and these are heart rate in beats per minute, heart rate variability (R - R variability), ectopic rate, ST segment depression, and the output signal of the Kohonen neural network analysis. Each of

41

these five different measurements are a significant
indicator of heart stress and these values can be
weighted and combined to give an overall level of heart
stress. The actual weighting applied to these signals
are application dependent. The requisite weighting can
be calculated using a neural network such as the multi-
layered perception. The operation of this network is
also described in the book by J Dayhoff.

The ability to monitor the general level of stress
experienced by the heart will enable a graded output
vector of heart stress. This enables the system to have
wide ranging applications.

Referring now to Figure 24, this is a schematic
drawing of a seventh embodiment of the present invention
wherein all the features of the signal processing and
analysis are carried out within a single unit which can
be a portable unit. The electrocardiograph signal from
the electrodes 100 is passed to an instrumentation
amplifier 200 and thence to an analogue to digital
converter 300. The digitised signal is then passed to
the preprocessor 400 for preprocessing the signal. The
preprocessed signal is then passed to the neural network
500 for analysis as hereinbefore described. The output
of the neural network is then output to an output device
600 which can comprise a display and/or audible output
unit. The output device can also trigger the storage of
any of the electrocardiograph signals.

42

Although the present invention has been described
with reference to specific embodiments, it should be
appreciated that the present invention is not limited to
such embodiments and it should be recognised that various
5   modifications are possible within the scope of the
invention claimed. For instance, the visual or audible
output can indicate the level of stress experienced by
the heart, or could more specifically provide a warning
when specific heart conditions are detected or an
10   undesirable change in the operation of the heart is
detected for a patient. Also, the detection of a
particular stress level, the detection of a change in
operation of the heart, or the detection of a specific
heart condition by the neural network can be used to
15   trigger the recording of the electrocardiograph trace for
analysis by a suitably trained person. Also, the number
of times that such events occur can be logged and
additionally the patient can be provided with an
indicated device to log a point in time if and when they
20   feel as well. This will provide additional information
to allow the identification of heart conditions.
Further, although reference has been made throughout to
the use of the Kohonen neural network, the present
invention is applicable to any equivalent neural network
25   which can provide for the analysis of the
electrocardiograph derived values.

43

<u>CLAIMS</u>

1.   Heart monitoring apparatus comprising:

       detection means for obtaining an electro-cardiograph

5   signal from a patient during a monitoring phase,

       preprocessing   means   for   processing   said

electrocardiograph signal to enhance the salient features

of the electrocardiograph signal and suppress the noise,

and to generate a plurality n of values representative

10   of the features of said electrocardiograph signal,

       storage means for storing a plurality m of n

dimensional reference vectors;

       Kohonen neural means for receiving said plurality

of values during the monitoring phase, for forming an n

15   dimensional vector from said plurality of values and for

comparing said n dimensional vector with said stored

plurality m of n dimensional reference vectors defining

an n dimensional Kohonen feature map to determine the

proximity of said n dimensional vector to said reference

20   vectors, and

       output means for outputting a signal if said Kohonen

neural network means determines that said n dimensional

vector is within or beyond a threshold range of said

reference vectors.

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2.   Heart monitoring apparatus as claimed in claim 1

wherein said Kohonen neural network means is adapted to

44

receive, during a learning phase, a plurality of reference values representative of the features of a reference electrocardiograph signal, to generate an n dimensional Kohonen feature map having said plurality m of reference vectors, and to store said reference vectors in said storage means.

3.   Heart monitoring apparatus as claimed in claim 2 wherein said detection means is adapted to obtain the reference electrocardiograph signal from the patient during the learning phase; said Kohonen neural network means is adapted to, during the monitoring phase, determine whether the n dimensional vector lies outside a predetermined range of said reference vectors; and said output means is adapted to output said signal if said Kohonen neural network means determines that said n dimensional vector lies outside said predetermined range.

4.   A heart monitoring apparatus as claimed in claim 2 including reference input means to input said reference electrocardiograph signal to said preprocessing means.

5.   Heart monitoring apparatus as claimed in claim 4 wherein input means is adapted to input said reference electrocardiograph signal representing normal range of operation of the heart; said Kohonen neural network means is adapted to determine whether the n dimensional vector

45

lies outside a predetermined range of said reference vectors; and said output means is adapted to output said signal if said Kohonen neural network means determines that said n dimensional vector lies outside said predetermined range.

6.   Heart monitoring apparatus as claimed in claim 4 or claim 5 wherein said input means is adapted to input at least one abnormal said reference electrocardiograph signal representing at least one known heart condition; said Kohonen neural network means is adapted to determine whether the n dimensional vector lies within a predetermined range of the abnormal reference vectors; and said output means is adapted to output said signal if said Kohonen neural network means determines that said n dimensional vector lies inside said predetermined range.

7.   Heart monitoring apparatus as claimed in claim 1 including vector inputting means for inputting said plurality m of reference vectors defining said n dimensional Kohonen feature map to said storage means.

8.   Heart monitoring apparatus as claimed in any preceding claim wherein said preprocessing means is adapted to carry out feature extraction to extract important features from said electrocardiograph signal,

46

and to use said features to form said plurality n of values.

9.   Heart monitoring apparatus as claimed in claim 8 wherein said preprocessing means is adapted to identify the peaks in said electrocardiograph signal and to measure the peak values to form said plurality n of values.

10. Heart monitoring apparatus as claimed in claim 9 wherein said preprocessing means is adapted to identify points midway between peaks and to measure the values of the electrocardiograph signal at said points to include the values in said plurality n of values.

11. Heart monitoring apparatus as claimed in claim 10 wherein said preprocessing means is adapted to identify the P, Q, R, S and T peaks and mid points between the P and Q peaks and the S and T peaks in said electrocardiograph signal, and to use the values of the signal at the identified peaks and points to form said plurality n of values.

12. Heart monitoring apparatus as claimed in any one of claims 1 to 7 wherein said preprocessing means is adapted to perform a transform of said electrocardiograph signal.

47

13. Heart monitoring apparatus as claimed in claim 12 wherein said preprocessing means is adapted to perform a Fourier transform and/or a Wavelett transform of said electrocardiograph signal.

14. Heart monitoring apparatus as claimed in any preceding claim wherein said Kohonen neural network means is adapted to carry out said comparison and determination by determining the differences between said reference vectors and said n dimensional vector for each dimension, summing the individual dimensional differences and comparing the sum with a threshold difference value.

15. Heart monitoring apparatus as claimed in any one of claims 1 to 13 wherein said Kohonen neural network means is adapted to carry out said comparison and determination by mapping said n dimensional vector into said Kohonen feature map, calculating the differences therebetween by Pythagoras' law, and comparing said differences with at least one difference threshold.

16. Heart monitoring apparatus as claimed in any preceding claim including filter means to remove distinctive irregular heartbeats from the electrocardiograph signal.

17. Heart monitoring apparatus as claimed in claim 16

48

including indicating means to indicate the occurrence of
the distinctive irregular beats.

18.   Heart monitoring apparatus as claimed in claim 16
or claim 17 wherein said filter means comprises a high
pass filter.

19.   Heart monitoring apparatus as claimed in claim 17
wherein said indicating means comprises a low pass
filter.

20.   Heart monitoring apparatus as claimed in any one of
claims 16 to 19 including averaging means for averaging
a plurality of said electrocardiograph signals or said
plurality n of values after removal of said distinctive
irregular beats by said filter means.

21.   Heart monitoring apparatus as claimed in claim 16
or claim 17 wherein said storage means stores a first set
of n dimensional reference vectors for the identification
of the distinctive irregular heartbeat, and a second set
of n dimensional reference vectors for monitoring regular
heartbeats;

         said Kohonen neural network means being adapted to,
during the monitoring phase, initially compare the n
dimensional vector with the stored first set of reference
vectors to identify the distinctive irregular heartbeats,

49

and to subsequently compare an n dimensional vector formed from regular heartbeats which exclude the distinctive irregular heartbeats with the second set of n dimensional reference vectors; and

5   said output means being adapted to output said signal if said Kohonen network means determines that said n dimensional vector formed from said regular heartbeats is within or beyond a threshold range of said second set of reference vectors.

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22.   Heart monitoring apparatus as claimed in claim 21 wherein said output means is adapted to output a signal indicating the occurrence of a distinctive irregular heartbeat when said Kohonen neural network means

15   determines that the n dimensional vector formed from the distinctive irregular heartbeat is within or beyond a threshold range of said first set of reference vectors.

23.   Heart monitoring apparatus as claimed in claim 22

20   including averaging means responsive to the signal indicating the occurrence of a distinctive irregular heartbeat for averaging a plurality of said electrocardiograph signals or said plurality n of values for a plurality of regular heartbeats.

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24.   Heart monitoring apparatus as claimed in claim 23 including normalising means for normalising the averaged

50

plurality of values by reducing the values by the mean
values and dividing by the standard deviation.

25. Heart monitoring apparatus as claimed in any
preceding claim including alarm generating means for
generating an alarm in response to the output signal of
said output means.

26. Heart monitoring apparatus as claimed in any
preceding claim including data storage means for storing
the electrocardiograph signal and/or the plurality of
values over a period of time in response to the output
signal of said output means.

27. Heart monitoring apparatus as claimed in any one of
claims 1 to 24 including postprocessing means to further
process the output signal of said output means, wherein
the output signal comprises an error vector, together
with further data related to heart operation to provide
an indication of heart stress.

28. Heart monitoring apparatus as claimed in claim 27
including means for providing said further data
comprising at least one of heart rate, heart rate
variation, rate of occurrence of distinctive irregular
heartbeats such as ectopic beats, and the difference
between the electrocardiograph signal value midway

51

between the S and T peaks and the electrocardiograph
signal value midway between the P and Q peaks.

29. Heart monitoring apparatus as claimed in claim 27
or claim 28 wherein said post processing means includes
neural network means.

30. Heart monitoring apparatus as claimed in claim 29
wherein said neural network means comprise multi-layered
perception means.

31. Heart monitoring apparatus as claimed in any
preceding claim including digitizing means for digitizing
said electrocardiograph signal for input to said
preprocessing means.

32. Heart monitoring apparatus as claimed in any
preceding claim wherein said detection means and said
output means are remote from one another, the apparatus
including communication means for airborne communications
therebetween.

33. Heart monitoring apparatus as claimed in claim 32
wherein said detection means is provided in a portable
housing arranged for attachment to the chest of said
patient, said portable housing including transmitter
means for transmitting signals to said output means; said

52

output means being provided in a base station, said base station including receiver means to receive the transmitted signals.

34.   Heart monitoring apparatus as claimed in claim 32 wherein said detection means comprises electrodes for attachment to the chest of a patient, said apparatus including a portable housing adapted to be carried by the patient, said portable housing including transmitter means for receiving signals from said electrodes and for transmitting signals to said output means; said electrodes being connected to said housing by wires; said output means being provided in a base station, said base station including receiver means to receive the transmitted signals.

35.   Heart monitoring apparatus as claimed in claim 33 or claim 34 wherein said transmission means and said reception means are arranged to respectively transmit and receive said electrocardiograph signals, said base station including said preprocessing means, and said Kohonen neural network means.

36.   Heart monitoring apparatus as claimed in claim 33 or claim 34 wherein said portable housing contains said preprocessing means, and said Kohonen neural network means; and said transmission and said reception means are

53

arranged to respectively transmit and receive signals
indicative of the proximity of said n dimensional vector
to said reference vectors.

5   37. Heart monitoring apparatus as claimed in any
preceding claim including storage means for storage of
the electrocardiograph signal or said plurality n of
values for a predetermined period of time in response to
the signal output by said output means.

10

38. Heart monitoring apparatus as claimed in claim 2
wherein said Kohonen neural network means is adapted to,
in the learning phase, receive a second plurality of
reference values representative of features of an
15   electrocardiograph signal indicative of a specific heart
condition, to generate a plurality of abnormal reference
vectors in said n dimensional Kohonen feature map; and
in the monitoring phase, to compare said n dimensional
vector with said plurality of abnormal reference vectors
20   to determine whether said n dimensional vector lies
within a predetermined range of said abnormal reference
vectors, said apparatus including heart condition warning
means responsive to said output signal for generating a
warning that said patient has said specific heart
25   condition if said Kohonen neural network means determines
that said n dimensional vector lies within said
predetermined range of said abnormal reference vectors.

54

39. A heart monitoring method comprising the steps of:

obtaining an electrocardiograph signal from a patient during a monitoring phase,

preprocessing the electrocardiograph signal to enhance the salient features of the electrocardiograph signal and suppress the noise, and to generate a plurality n of values representative of the features of the electrocardiograph signal,

forming an n dimensional vector from said plurality n of values,

comparing the n dimensional vector with a stored plurality m of n dimensional reference vectors defining an n dimensional Kohonen feature map to determine the proximity of the n dimensional vector to said reference vectors, and

outputting a signal if it is determined that the n dimensional vector is within or beyond a threshold range of said reference vectors.

40. A heart monitoring method as claimed in claim 39 including the steps during a learning phase of:

receiving a plurality of reference values representative of features of a reference electrocardiograph signal,

generating an n dimensional Kohonen feature map having said plurality m of reference vectors, and

storing said reference vectors.

55

41.   A heart monitoring method as claimed in claim 40
including the steps of:

obtaining said electrocardiograph signal from said
patient during the learning phase; and

5   during the monitoring phase determining whether the
n dimensional vector lies outside a predetermined range
of said reference vectors; and

outputting said signal if it is determined that said
n dimensional vector lies outside said predetermined

10   range.

42.   A heart monitoring method as claimed in claim 40
including the steps of:

during the learning phase, obtaining said reference

15   electrocardiograph signal representative of a normal
range of operation of the heart, and processing the
reference electrocardiograph signal to generate said
plurality of reference values, and

during the monitoring phase, determining whether the

20   n dimensional vector lies outside a predetermined range
of said reference vectors, and outputting said signal if
it is determined that said n dimensional vector lies
outside said predetermined range.

25   43.   A heart monitoring method as claimed in claim 41 or
claim 42 including the steps of:

during the learning phase, inputting at least one

56

abnormal said reference electrocardiograph signal representing at least one known heart condition; and

during the monitoring phase, determining whether the n dimensional vector lies within a predetermined range of the abnormal reference vectors, and outputting said signal if it is determined that said n dimensional vector lies inside said predetermined range.

44. A heart monitoring method as claimed in claim 39 including the step of inputting and storing said plurality m of reference vectors defining said n dimensional Kohonen feature map.

45. A heart monitoring method as claimed in any one of claims 39 to 44 wherein the preprocessing step comprises the steps of extracting important features from said electrocardiograph signal, and generating said plurality n of values from the extracted features.

46. A heart monitoring method as claimed in claim 45 wherein the processing step includes the steps of identifying the peaks in said electrocardiograph signal, and measuring the peak values to form said plurality n of values.

47. A heart monitoring method as claimed in claim 46 wherein said preprocessing step includes the steps of

57

identifying points midway between peaks, and measuring
the values of said electrocardiograph signal at said
points to include the values in said plurality n of
values.

48. A heart monitoring method as claimed in claim 47
wherein said preprocessing step includes the steps of
identifying the P, Q, R, S and T peaks and mid points
between the P and Q and the S and T peaks in said
electrocardiograph signal, and using the values for the
electrocardiograph signal at the identified peaks and
points to form said plurality n of values.

49. A heart monitoring method as claimed in any one of
claims 39 to 45 wherein said preprocessing step includes
the step of performing a transform of said
electrocardiograph signal.

50. A heart monitoring method as claimed in claim 49
wherein said preprocessing step includes the step of
performing a Fourier transform and/or a Wavelett
transform of said electrocardiograph signal.

51. A heart monitoring method as claimed in any one of
claims 39 to 50 wherein the comparing step includes the
step of determining the differences between said
reference vectors and said n dimensional vector for each

58

dimension, summing the individual dimensional differences, and comparing the sum with a threshold difference value.

52. A heart monitoring method as claimed in any one of claims 39 to 50 wherein the comparing step includes the steps of mapping said n dimensional vector into said Kohonen feature map, calculating the differences therebetween by Pythagoras' law, and comparing said differences with at least one difference threshold.

53. A heart monitoring method as claimed in any one of claims 39 to 52 including the step of removing distinctive irregular heartbeats from said electrocardiograph signal.

54. A heart monitoring method as claimed in claim 53 including the step of indicating the occurrence of the distinctive irregular beats.

55. A heart monitoring method as claimed in claim 53 or claim 54 including the step of averaging a plurality of said electrocardiograph signals or said plurality of values after removal of said distinctive irregular heartbeats.

56. A heart monitoring method as claimed in claim 53 or

59

claim 54 including the steps of during the learning phase, storing a first set of n dimensional reference vectors for the identification of the distinctive irregular heartbeat, and a second set of n dimensional reference vectors for monitoring regular heartbeats;

the comparing step comprising initially comparing the n dimensional vector with the stored first set of reference vectors to identify the distinctive irregular heartbeats, and subsequently comparing an n dimensional vector formed from regular heartbeats excluding the distinctive irregular heartbeats with said second set of n dimensional reference vectors; and

the outputting step comprising outputting said signal if it is determined that said n dimensional vector formed from said regular heartbeats is within or beyond a threshold range of said second set of reference vectors.

57. A heart monitoring method as claimed in claim 56 wherein the outputting step includes the step of outputting a signal indicating the occurrence of a distinctive irregular heartbeat when it is determined that the n dimensional vector formed from the distinctive irregular heartbeat is within or beyond a threshold range of said first set of reference vectors.

58. A heart monitoring method as claimed in claim 57

60

including the step of averaging a plurality of said electrocardiograph signals or said plurality n of values for a plurality of regular heartbeats.

5   59. A heart monitoring method as claimed in claim 58 including the step of normalising the averaged plurality of values by reducing the values by the mean values and dividing by the standard deviation.

10   60. A heart monitoring method as claimed in any one of claims 39 to 59 including the step of generating an alarm in response to the output signal.

61. A heart monitoring method as claimed in any one of
15   claims 39 to 60 including the step of storing the electrocardiograph signal and/or the plurality of values over a period of time in response to the output signal.

62. A heart monitoring method as claimed in any one of
20   claims 39 to 59 including the step of post processing the output signal, wherein the output signal comprises an error vector, together with further data related to heart operation to provide an indication of heart stress.

25   63. A heart monitoring method as claimed in claim 62 including the step of providing the further data comprising at least one of heart rate, heart rate

61

variation, rate of occurrence of distinctive irregular
heartbeats such as ectopic beats, and the difference
between the electrocardiograph signal value midway
between the S and T peaks and the electrocardiograph
signal value midway between the P and Q peaks.

64.   A heart monitoring method as claimed in claim 62 or
claim 63 wherein the post processing is carried out by
a neural network such as the multi-layered perception.

65.   A heart monitoring method as claimed in any one of
claims 39 to 64 indicating the step of digesting the
electrocardiograph signal prior to preprocessing.

66.   A heart monitoring method as claimed in any one of
claims 39 to 65 wherein the obtaining step is carried out
at one location and the output step is carried out at a
second location remote from said first location, the
method including providing airborne communications
between the two locations.

67.   A heart monitoring method as claimed in claim 66
wherein the electrocardiograph signals are obtained by
a portable unit attached to the patient, and the output
signal is provided at a base station remote from said
portable unit.

62

68. A heart monitoring method as claimed in claim 67 wherein the electrocardiograph signals are transmitted to said base station where said preprocessing step, forming step, comparing step and output step take place.

69. A heart monitoring method as claimed in claim 67 wherein the obtaining and preprocessing steps take place in said portable unit, and said plurality n of values are transmitted to said base station.

70. A heart monitoring method as claimed in any one of claims 39 to 69 including the step of storing the electrocardiograph signal and said plurality of values for a predetermined period of time in response to said output signal.

71. A heart monitoring method as claimed in claim 40 including the steps of:

in the learning phase, receiving a second plurality of reference values representative of features of an electrocardiograph signal indicative of a specific heart condition, and generating a plurality of abnormal reference vectors in said n dimensional Kohonen feature map; and

in the monitoring phase, comparing said n dimensional vector with said plurality of abnormal reference vectors to determine whether said n dimensional

63

vector lies within a predetermined range of said abnormal reference vectors, and generating a warning that said patient has said specific heart condition if it is determined that said n dimensional vector lies within

5   said predetermined range of said abnormal reference vectors.

72.   Heart monitoring apparatus comprising:

input means for receiving an electrocardiograph
10   signal from a patient;

preprocessing means for processing said electrocardiograph signal to suppress the noise and extract important features of the electrocardiograph signal to obtain an n dimensional vector comprising a
15   plurality n of values representative of the important features of said electrocardiograph signal;

storage means fro storing a plurality m of n dimensional reference vectors;

neural network means for receiving said n
20   dimensional vector, for comparing said n dimensional vector with said stored plurality m of n dimensional reference vectors defining an n dimensional volume to determine the proximity of said n dimensional vector to said n dimensional volume, and for outputting an
25   indication of whether said n dimensional vector lies within or beyond a threshold range of said n dimensional volume.

64

73.   Heart monitoring apparatus as claimed in claim 72 wherein said neural network means comprises Kohonen neural network means and said reference vectors form an n dimensional Kohonen feature map.

74.   Heart monitoring apparatus comprising:

measuring means for measuring the activity of a patient's heart and outputting an electrocardiograph signal during a learning phase and a monitoring phase;

preprocessing   means   for   processing   said electrocardiograph signal to extract important features therefrom to generate an n dimensional vector comprising a plurality n of values representative of the features of the electrocardiograph signal; and

Kohonen   neural   network   means   for,   during   the learning phase, using said n dimensional vector to generate an n dimensional Kohonen feature map with a plurality m of reference vectors defining a normal range of electrocardiograph features for said patient, and for during the monitoring phase, comparing said n dimensional vector with said plurality of m dimensional reference vectors defining the n dimensional Kohonen feature map to determine the proximity of said n dimensional vector to said reference vectors, and to output a signal if it is determined that said n dimensional vector is within or beyond a threshold range of said reference vectors.

Fig 1

2/24

Remote base station

400

Neural computer

Display and audible alarm

600

300

Receiver

500

100

200

Fig 2

3/24

electrodes → amp → AGC → Modulator

rf generator → Modulator

6 (antenna)

1 — electrodes
2 — amp
3 — AGC
4 — Modulator
5 — rf generator

7 (antenna) → demodulator → ADC → processor

processor: preproc → neural network → alarm

8 — demodulator
9 — ADC
10 — preproc
11 — neural network
12 — processor
13 — alarm

Fig 3

4/24

Fig 4

Fig 5

6/24

Fig 6

7/24

Fig 7

8/24

Fig 8

Pattern space

Fig 9a

Fig 9b

Pattern space

Fig 10

Pattern space
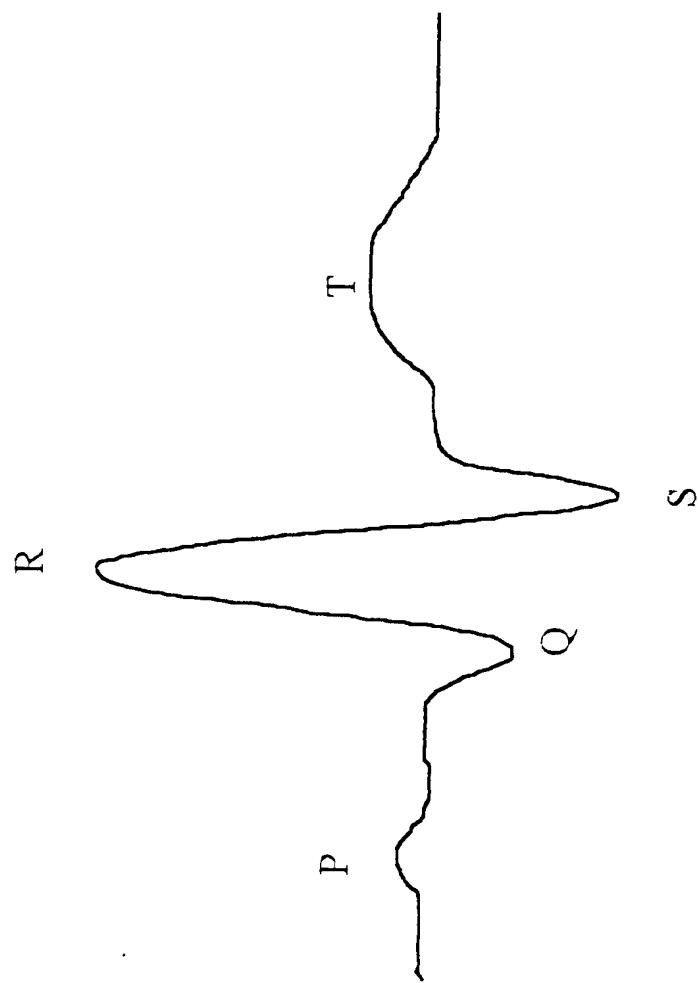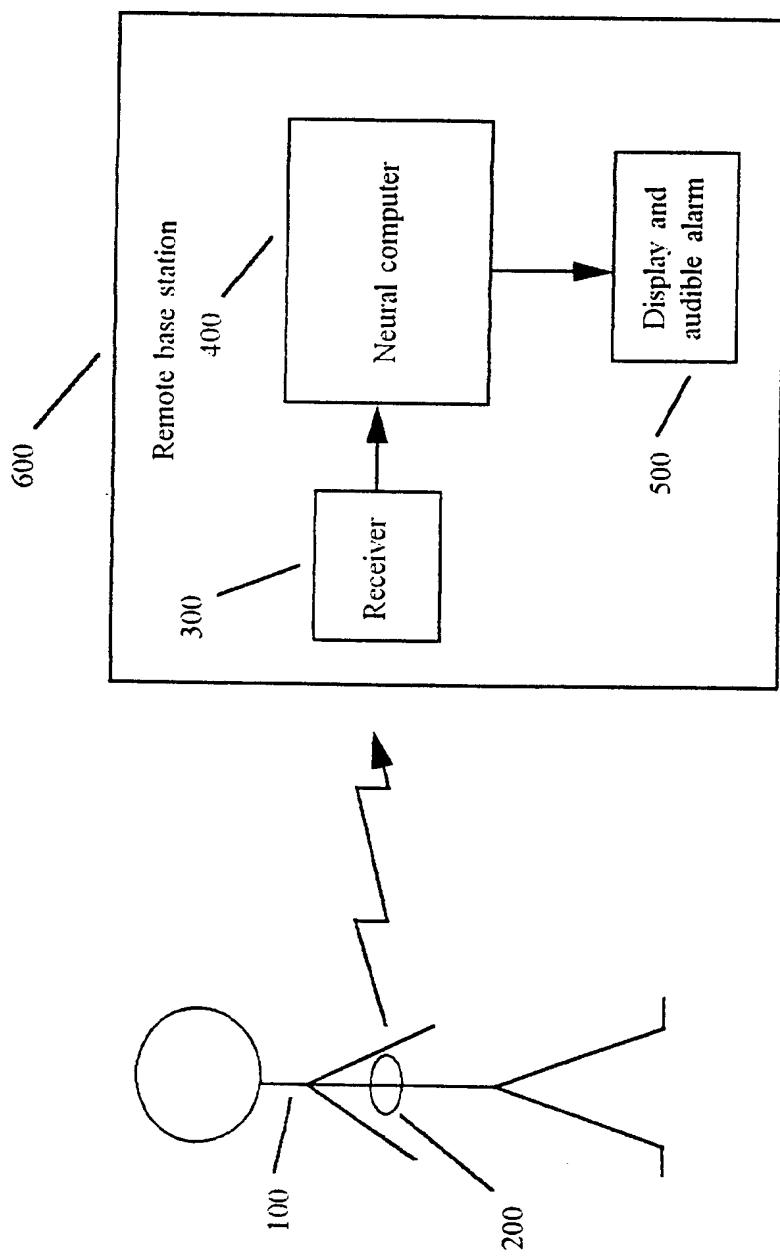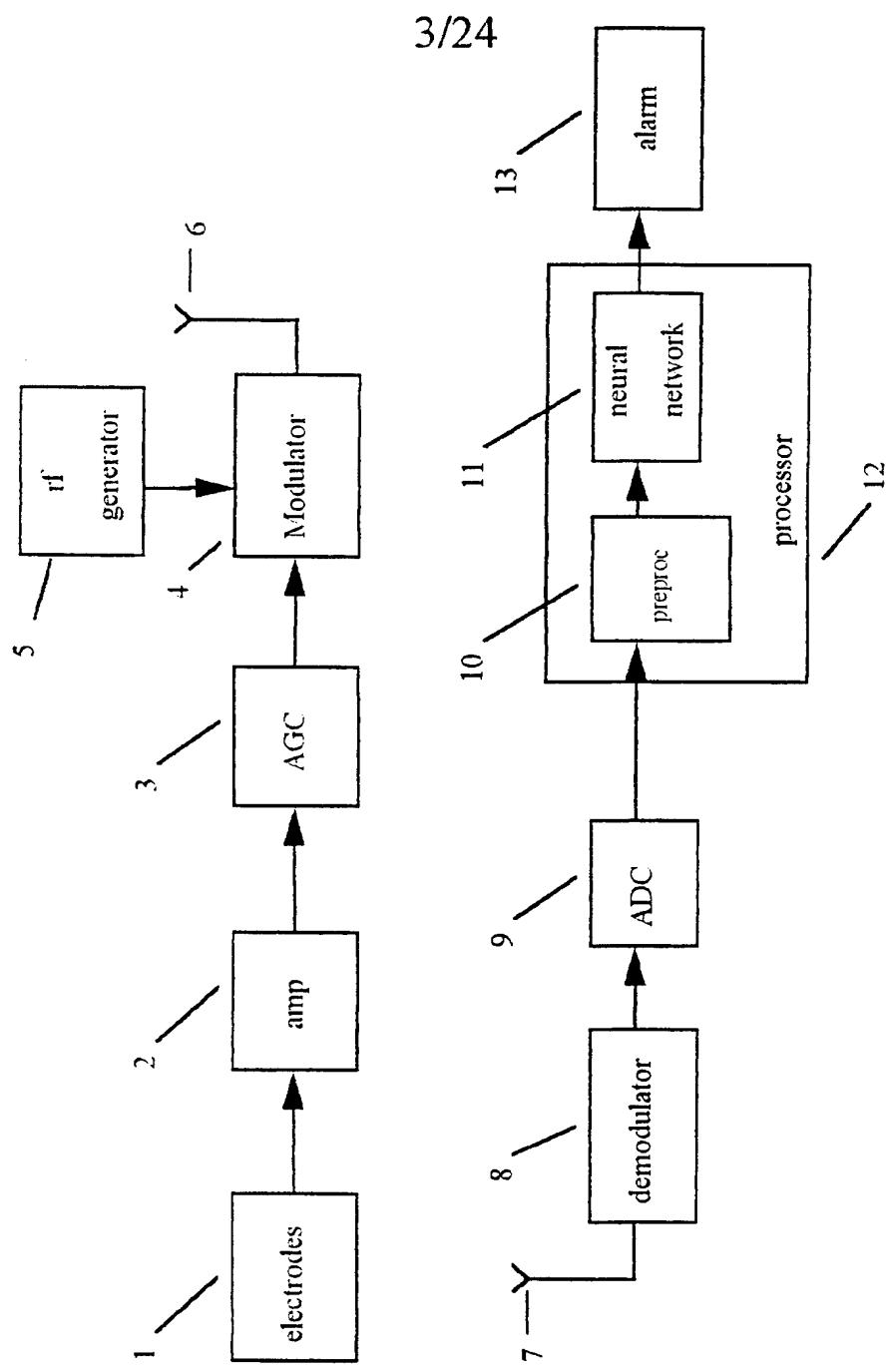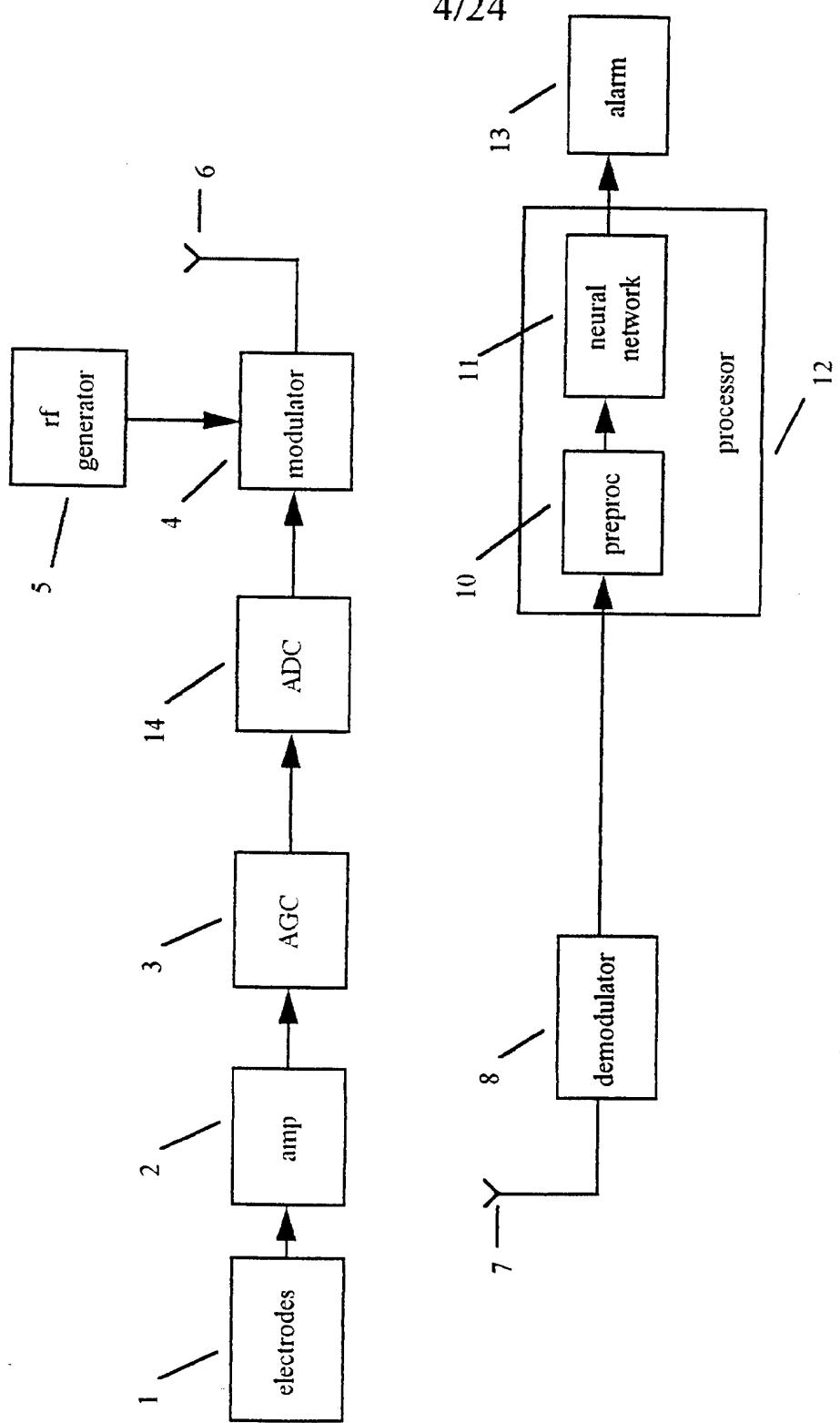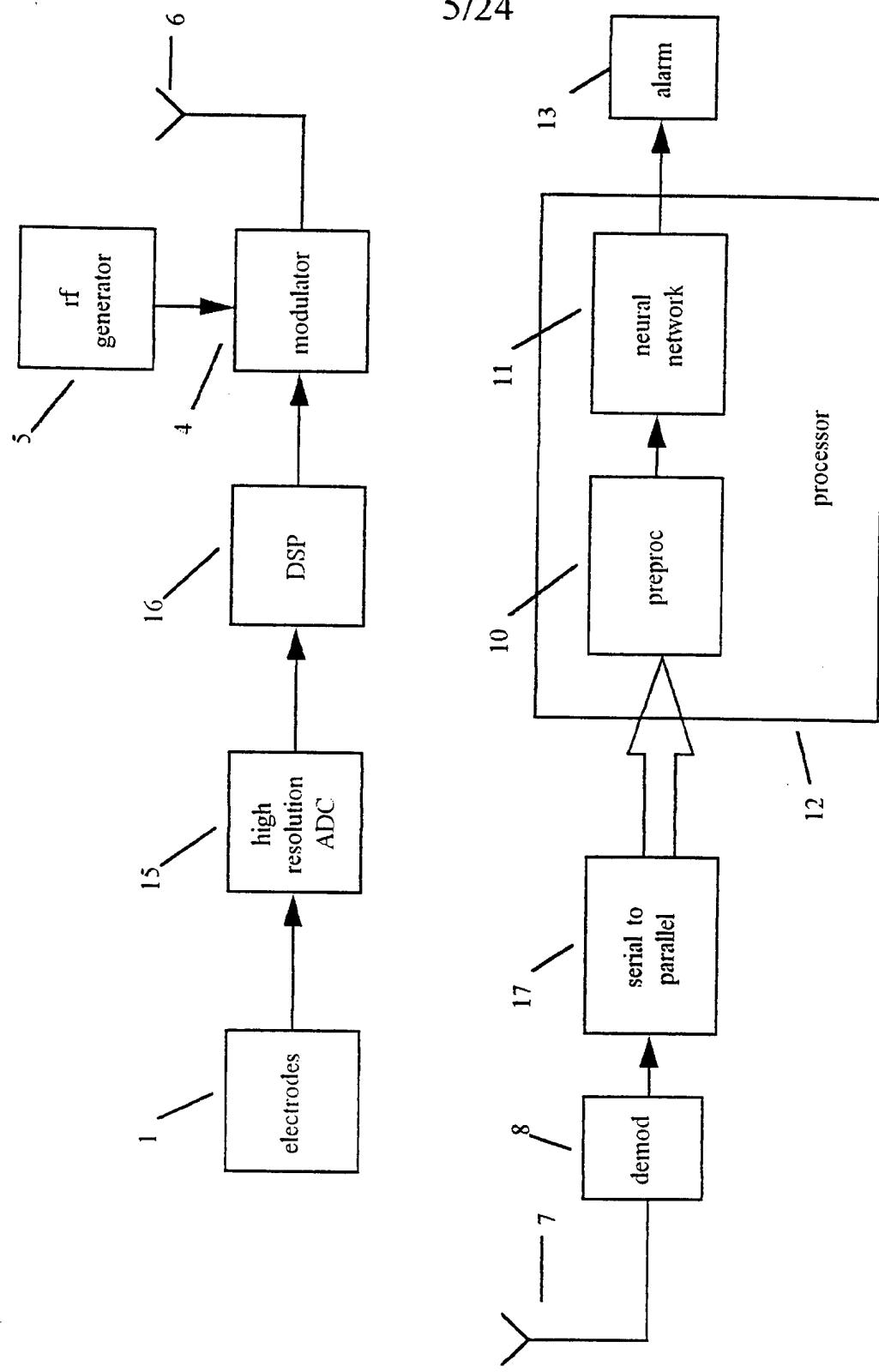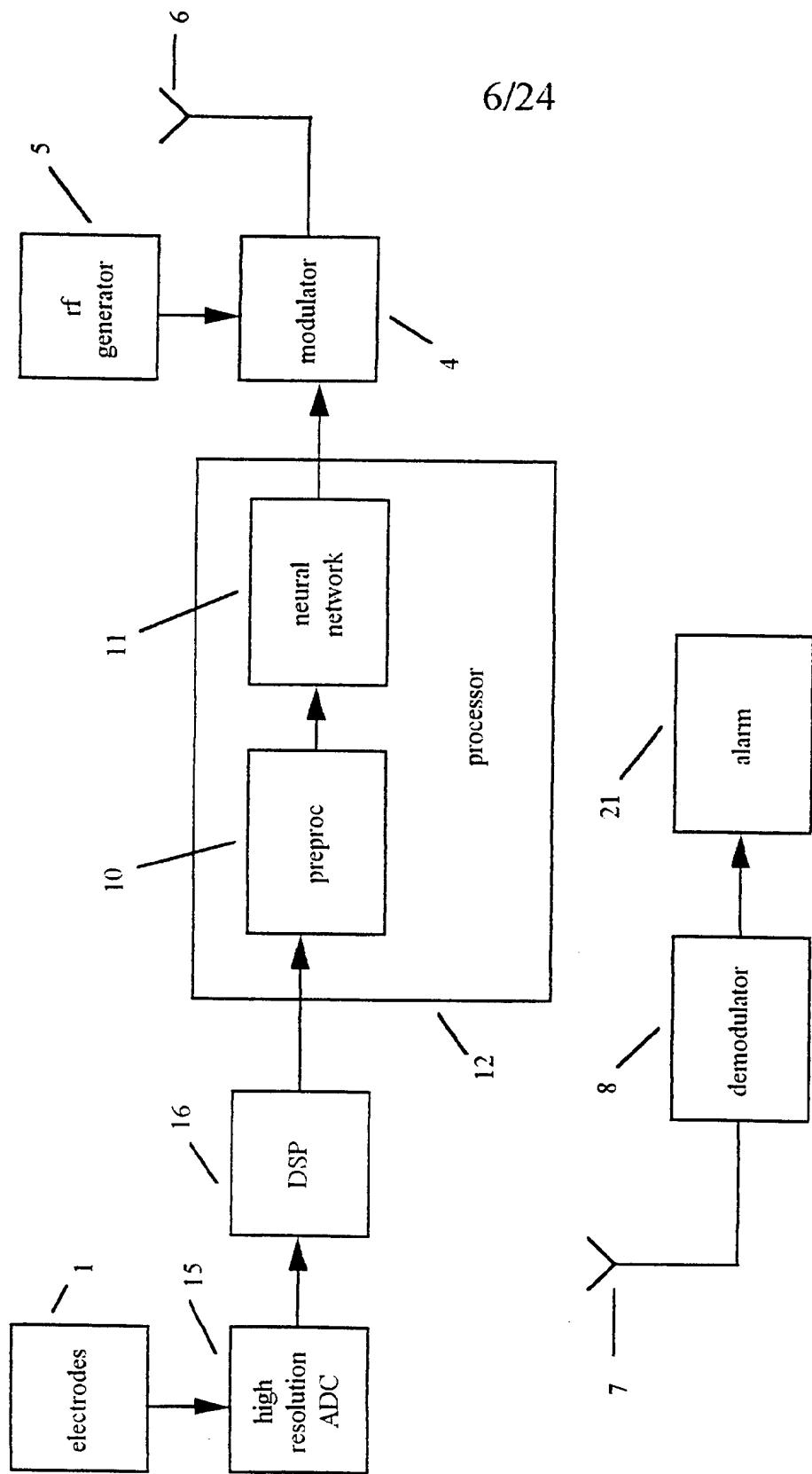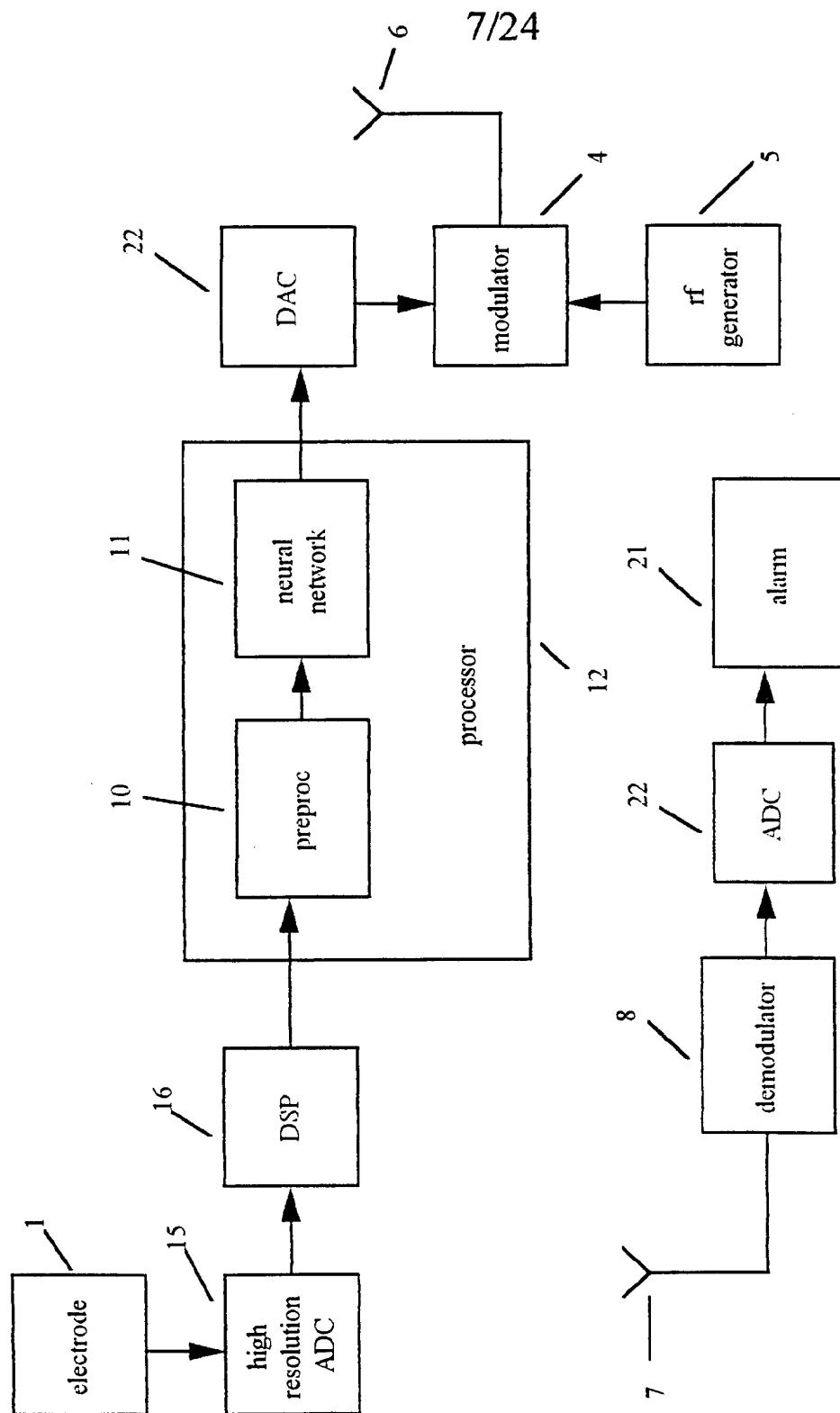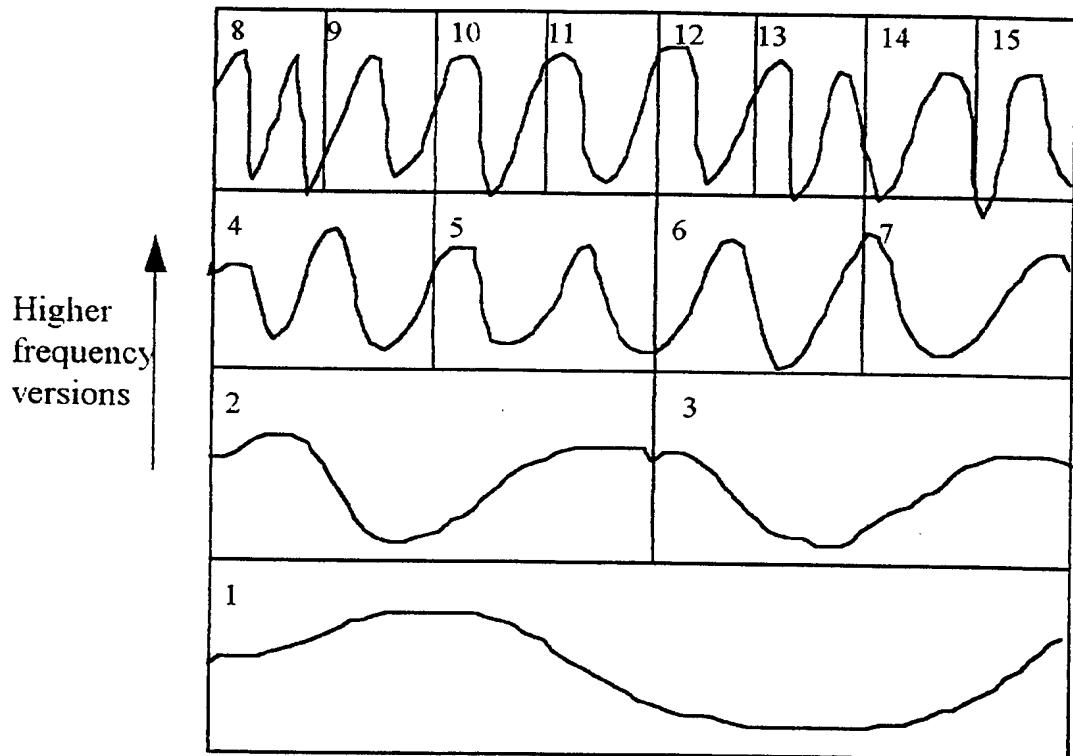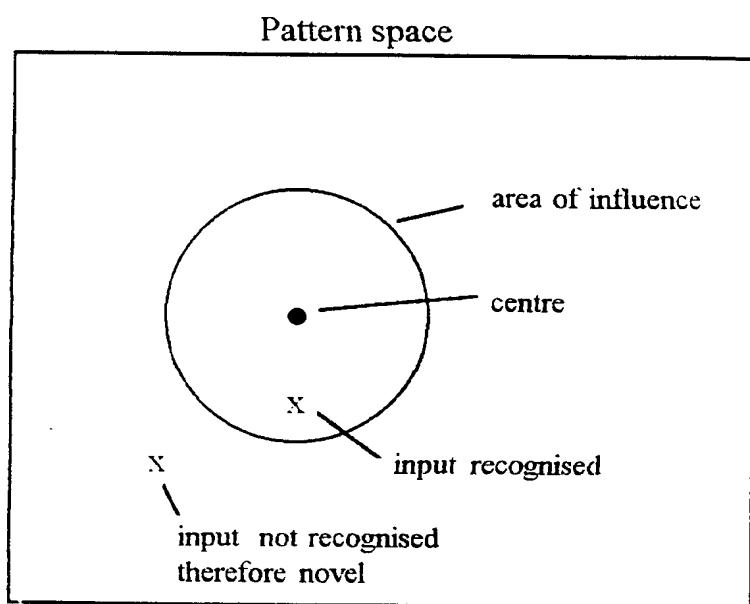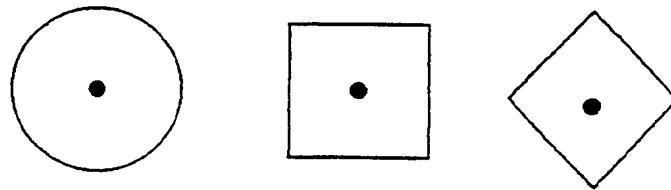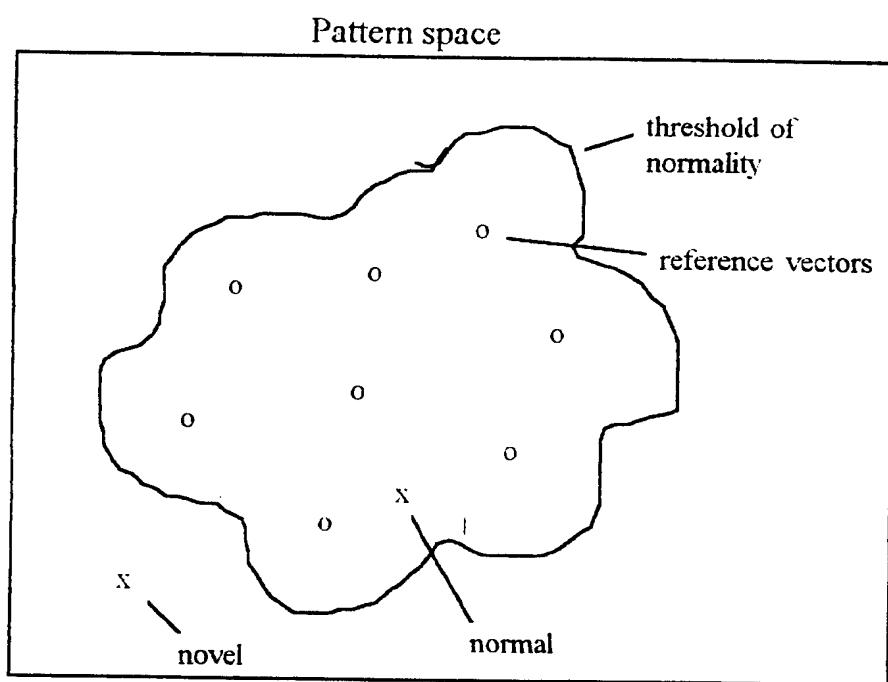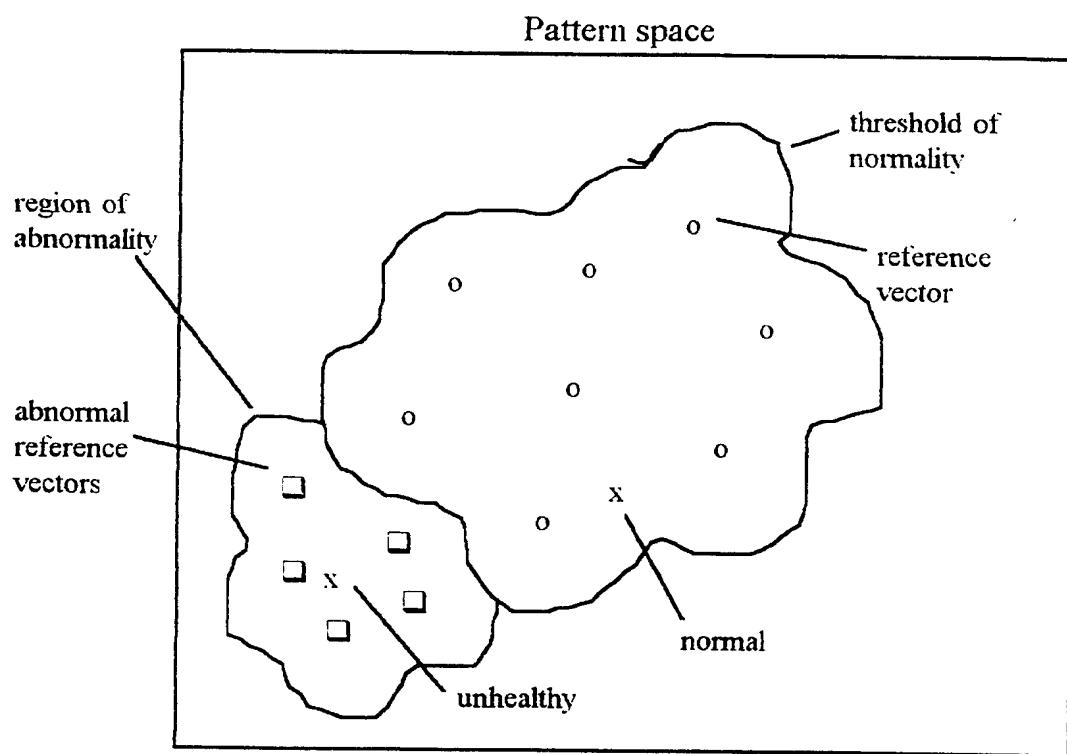
Fig 11
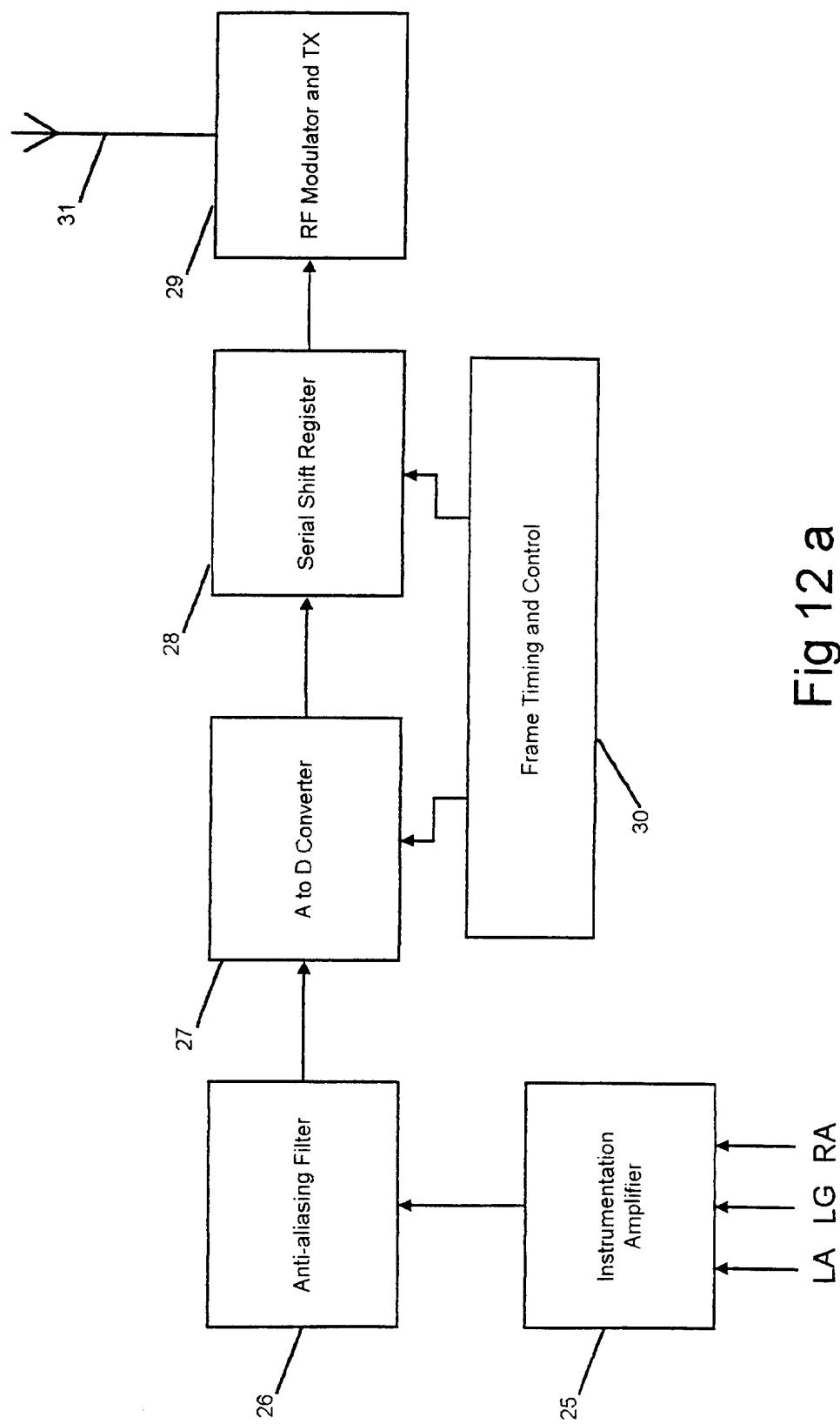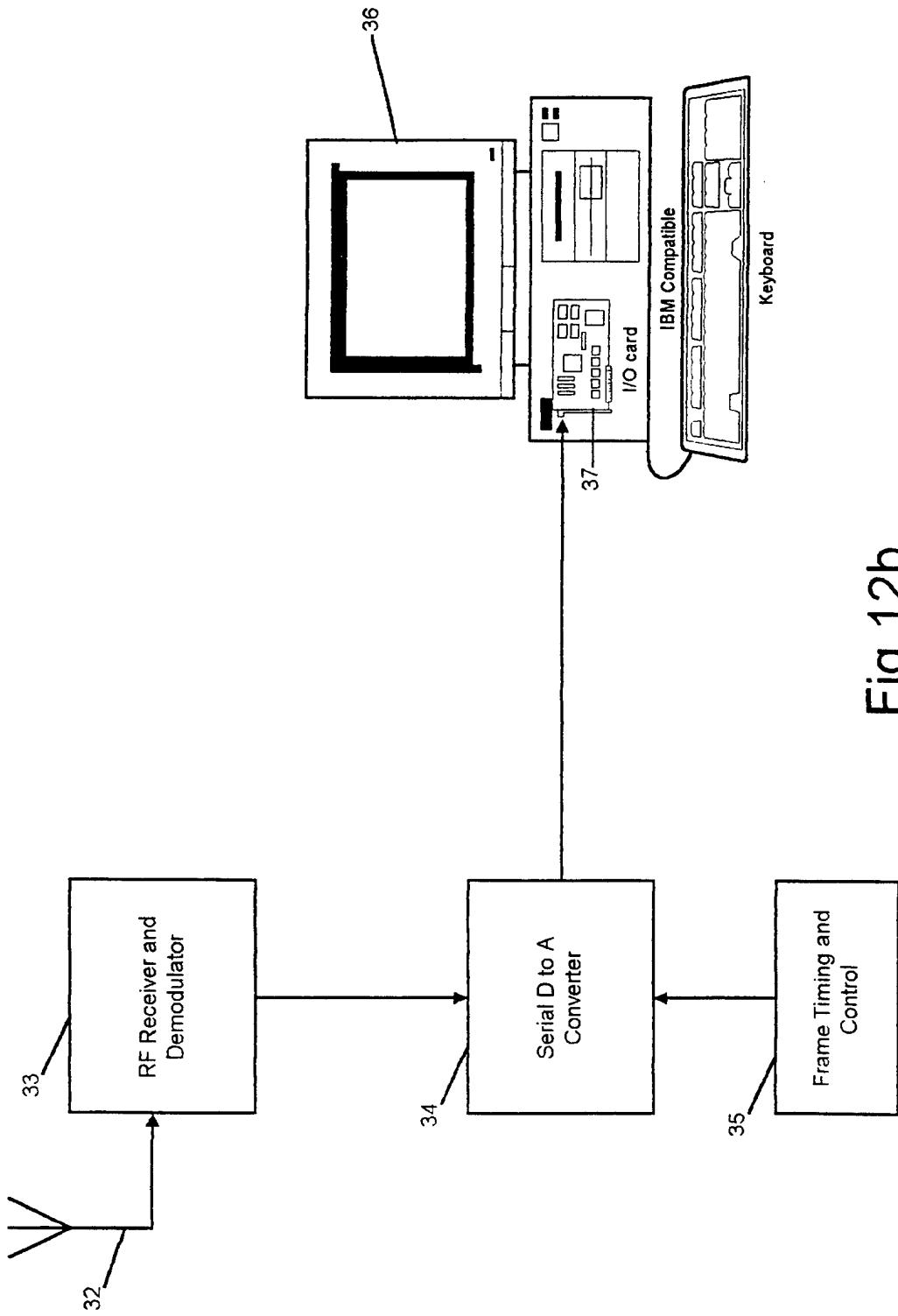
Fig 12 a

Fig 12b
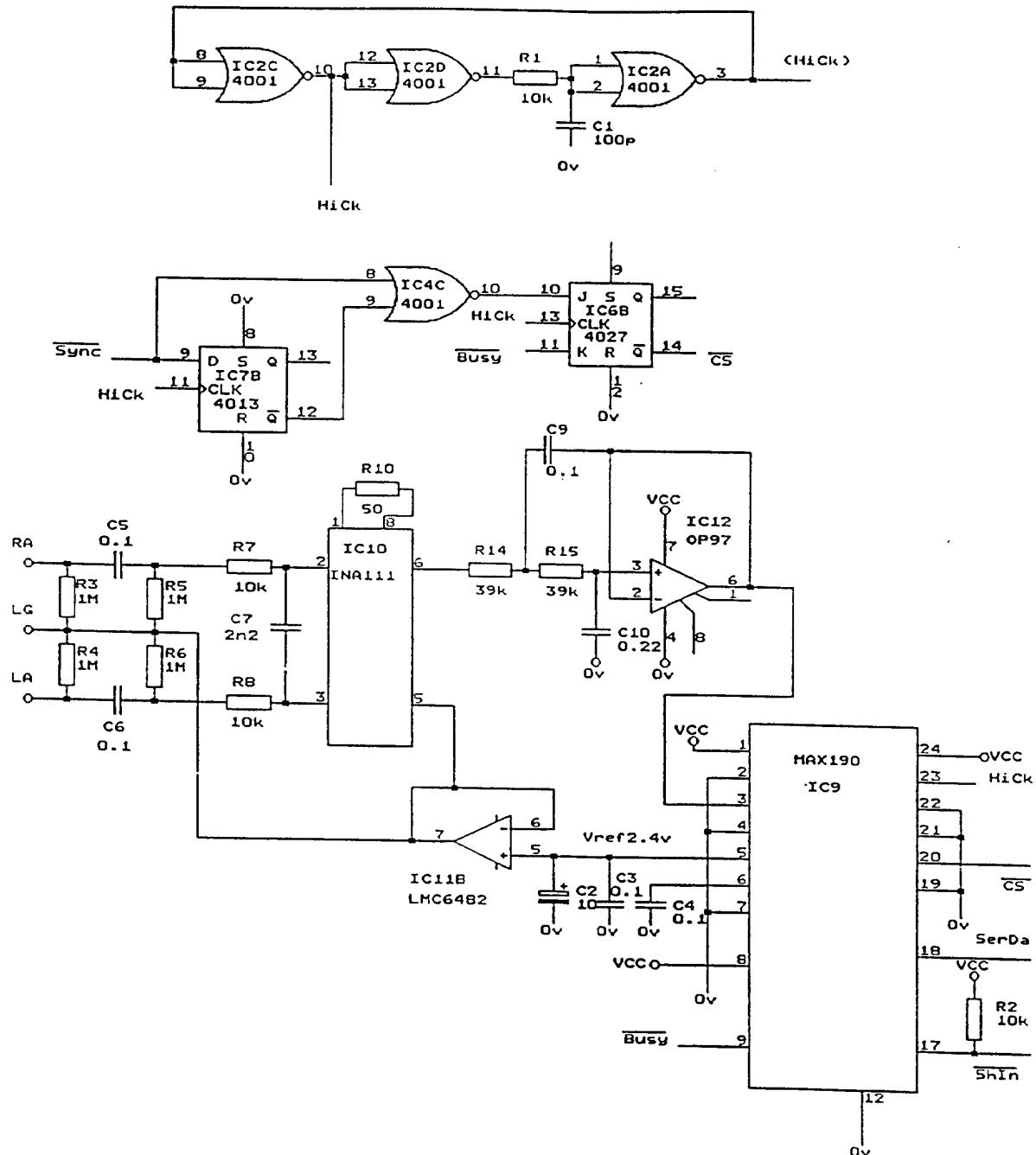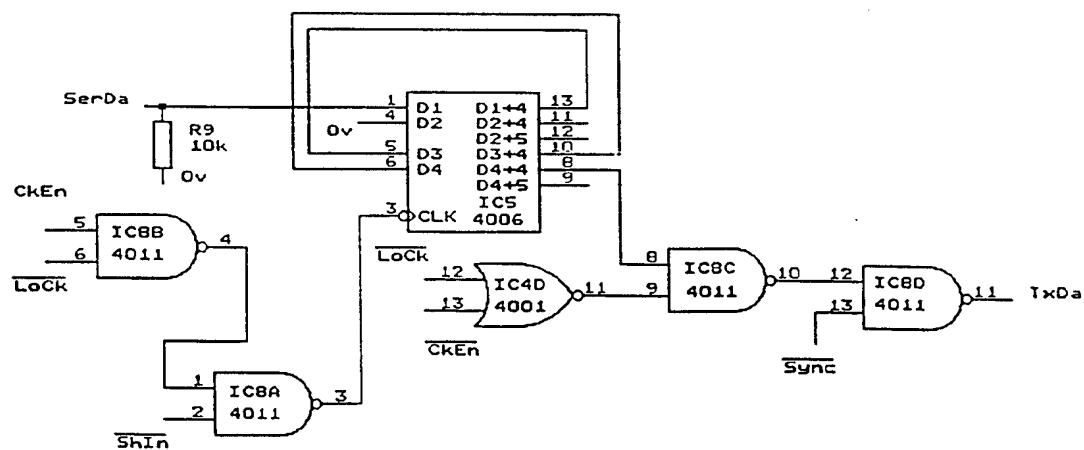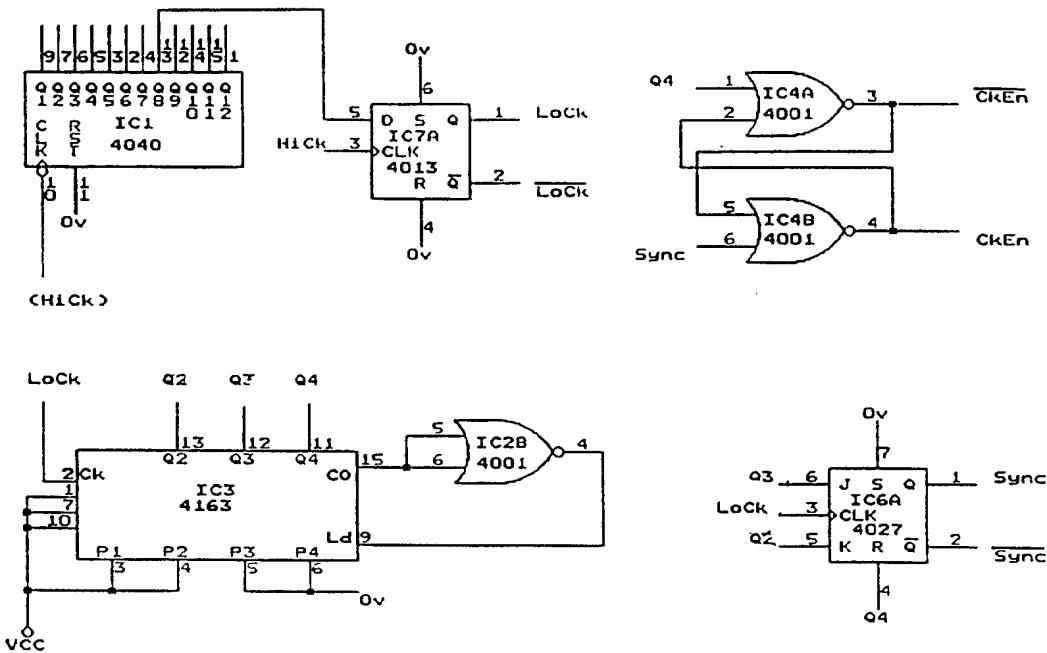
Fig 13a

Fig 13b
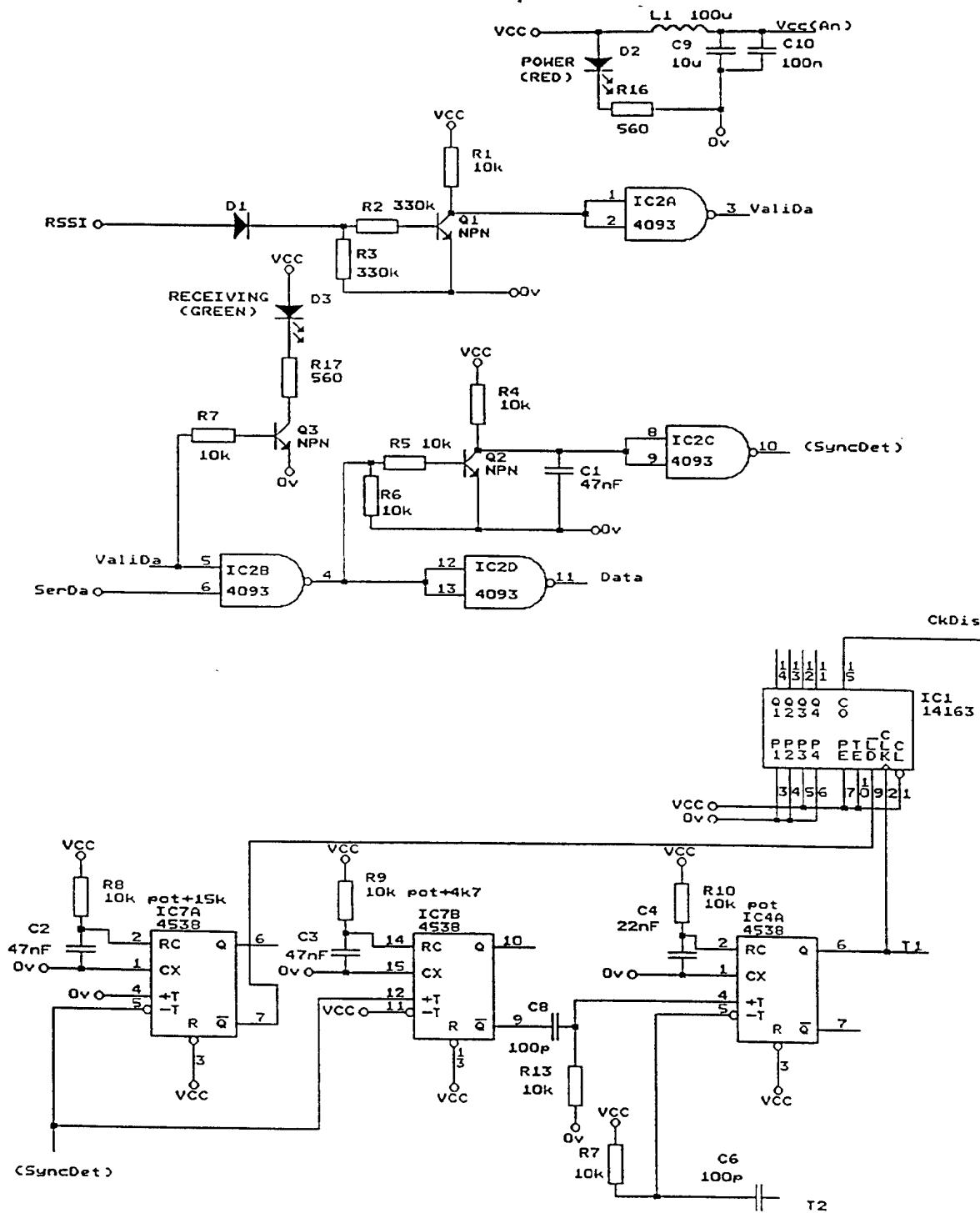
Fig 14a

17/24
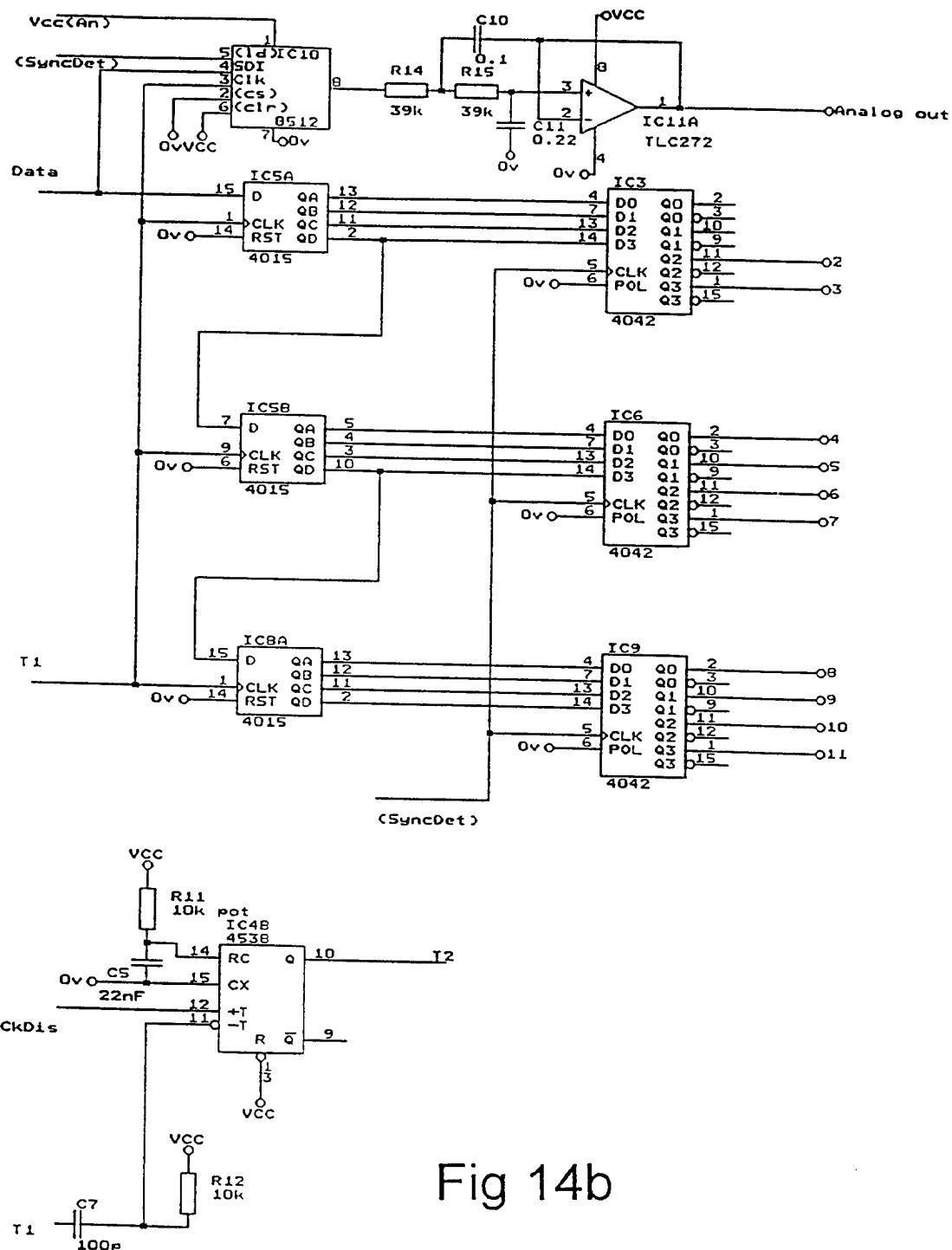

Fig 14b
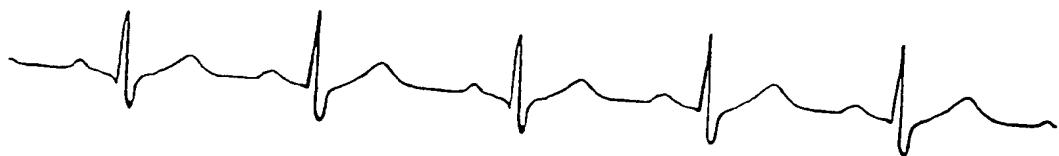
Fig 15
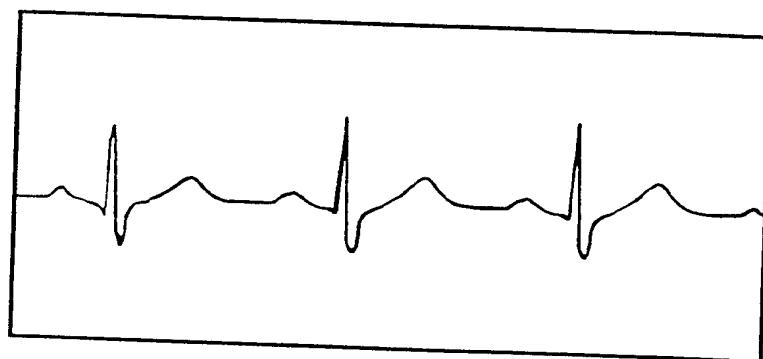
Fig 16
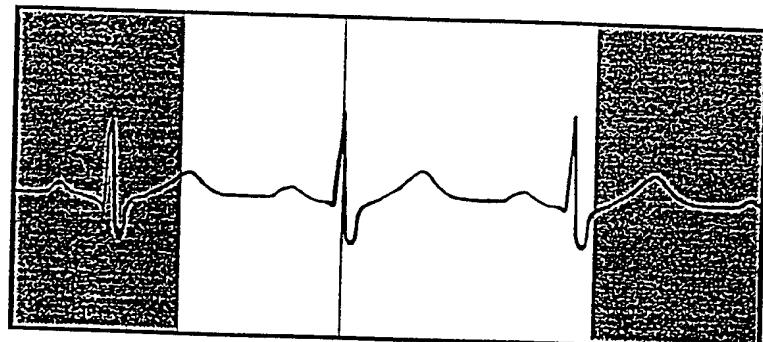
Fig 17
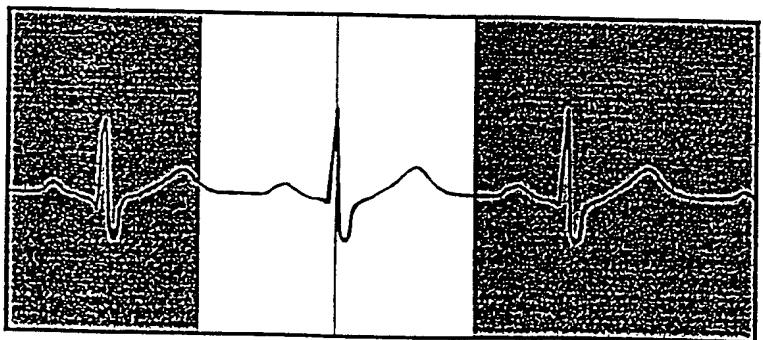
Fig 18

19/24
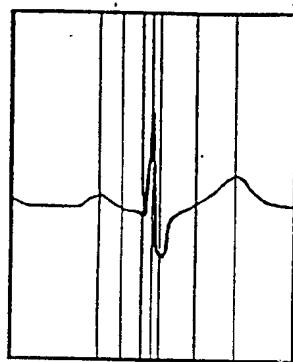
Fig 20a
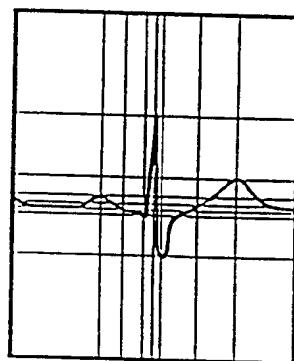
Fig 20b


Fig 20c

20/24

Fig 21

Fig 19

```
┌──────────────────────┐
│    Input ECG data    │  S1
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│  Data captured at a  │
│ sample rate of 1kHz  │  S2
│ and stored as a *.wav│
│         file         │
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│ Convert *.wav file to│  S3
│      Excel file      │
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│  Extract amplitude   │  S4
│     information      │
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│  Subsample at 100Hz  │  S7
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│   Take a 3 second    │  S6
│ window (300 samples) │
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│ Scan middle 2 seconds│  S7
│  for  and R wave peak│
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│   Create 1 second    │
│subwindow with R wave │  S8
│     centralised      │
└──────────────────────┘
            │
            ▼
┌──────────────────────┐
│ Normalise amplitudes │  S9
│  to range from 0 to 1│
└──────────────────────┘
            │
            ▼
         ( A )
```

(B)

Fig 22a

Fig 22b

| Locate T wave | S100 |

↓

| Measure amplitude of R and T | S110 |

↓

| Take the average of the R and T wave amplitudes | S120 |

↓

| Locate next or previous R wave in the 3 second window by looking for peak above the average | S130 |

Fig 23

↓

| Calculate and output heart rate | S140 |

↓

| Calculate and output R-R variability for the most recent 200 beats | S150 |

↓

| Calculate position of P Q and S using a priori knowledge of beat shape | S160 |

↓

| Calculate mid points PQ and ST | S170 |

↓

| Calculate the 12 values | S180 |

Fig 24

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
IPC 6   A61B5/0472   G06F19/00

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC 6   A61B   G06F

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	EP,A,0 433 626 (WAYNE STATE UNIVERSITY) 26 June 1991	1-9, 32-36, 39-46, 66-68, 72-74
A	see column 8, line 11 - column 24, line 31; tables 1-8	10-31, 37,38, 47-65, 70,71

	-/--	

[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

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"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
6 November 1996	2 0. 11. 96

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (- 31-70) 340-2040, Tx. 31 651 epo nl, Fax (- 31-70) 340-3016	Weihs, J

Form PCT ISA 210 (second sheet) (July 1992)

INTERNATIONAL SEARCH REPORT

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	PROCEEDINGS OF 1994 IEEE INTERNATIONAL CONFERENCE ON NEURAL NETWORKS , vol. 6, 27 June 1994 - 2 July 1994, ORLANDO, US, pages 3552-3558, XP000510479 TONI CONDE: "automatic neural detection of anomalies in electrocardiogram (ecg) signals" * whole document * ---	1-9, 39-46, 72-74
Y	US,A,3 832 994 (HAIM I.BICHER ET AL) 3 September 1974 see column 2, line 33 - line 64; table 1 ---	32-36, 66-68
A	DE,A,43 07 545 (SIEMENS AG) 15 September 1994 see column 3, line 23 - column 6, line 1; tables 1,2 -----	1,39,72, 74

2

INTERNATIONAL SEARCH REPORT

information on patent family members

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		JP-A-	3207342	10-09-91
US-A-3832994	03-09-74	NONE		
DE-A-4307545	15-09-94	NONE		